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Inhomogeneous Deformation of Brain Tissue During Tension Tests

Badar Rashid[a], Michel Destrade[b,a], Michael D Gilchrist[a*]

[a]*School of Mechanical and Materials Engineering, University College Dublin, Belfield, Dublin 4, Ireland*

[b]*School of Mathematics, Statistics and Applied Mathematics, National University of Ireland Galway, Galway, Ireland*

**Corresponding Author*

Tel: + 353 1 716 1884/1991, + 353 91 49 2344 Fax: + 353 1 283 0534

Email: Badar.Rashid@ucdconnect.ie (B. Rashid), michael.gilchrist@ucd.ie (M.D. Gilchrist), michel.destrade@nuigalway.ie (M. Destrade)

Abstract Mechanical characterization of brain tissue has been investigated extensively by various research groups over the past fifty years. These properties are particularly important for modelling Traumatic Brain Injury (TBI) by using finite element human head models to simulate brain injuries under different impact conditions. They are also increasingly important for computer assisted neurosurgery. During severe impact conditions, brain tissue experiences compression, tension and shear; however only limited tests have been performed in tension. Typically, cylindrical specimen are prepared and glued to platens to perform tensile tests which produce an inhomogeneous deformation field near the boundaries, thus contributing to higher magnitudes of stresses. In this research, we present the design and calibration of a High Rate Tension Device (HRTD) capable of performing tests up to a maximum strain rate of 90/s. We use experimental and numerical methods to investigate the effects of inhomogeneous deformation of porcine brain tissue during tension at different specimen thicknesses (4.0 – 14.0 mm), by performing tension tests at a strain rate of 30/s. One-term Ogden material parameters (μ = 4395.0 Pa, α = - 2.8) were derived by performing an inverse finite element analysis to model all experimental data. A similar procedure was adopted to determine the Young's modulus (E = 11200 Pa) of the linear elastic regime. Based on this analysis, brain specimens of aspect ratio (diameter/thickness) S = 10/10 or lower (10/12, 10/13) are considered suitable for minimizing the effects of inhomogeneous deformation during tension tests.

Keywords: Traumatic brain injury, TBI, Impact, Compression, Shear, Ogden, Hyperelastic, Elastic

1 Introduction

Procedures for the mechanical characterization of soft biological tissues (such as kidney, lungs, skin and brain tissue) at quasi-static loading have been well established over the past five decades. Almost all soft tissues are now considered to be nonlinear, anisotropic and viscoelastic in nature. Determining the mechanical parameters of soft biological tissues becomes a formidable challenge at high dynamic velocities. During a severe impact to the head, brain tissue experiences a mixture of compression, tension and shear. In order to investigate the mechanisms involved in Traumatic Brain Injury (TBI), several research groups have investigated the brain's mechanical properties over a wide range of loading conditions by adopting different test protocols [1-38]. However, few tests have been performed in tension [39-41] so far.

Moreover, *diffuse axonal injury* (DAI) is the most severe form of injury, occurring at shear strains of approximately 10% – 50% and strain rates of approximately 10 – 50/s [4-9]. Recently, Tamura [40] designed an apparatus to perform tests at 0.9, 4.3 and 25/s, but only the fastest of these rates is close to DAI impact speeds. The Kolsky test apparatus is usually used to perform compression tests at high strain rates, but it is more suitable for strain rates > 100/s. Based on the specific range of strain and strain rates which are injurious to axons during DAI, there is now an urgent need to develop tensile test equipment that can perform tests at strain rates up to 100 /s.

Tensile tests on engineering materials are typically performed using dog bone shaped test specimens to ensure homogeneous deformation over the required gauge length. However cylindrical specimens are more easily used for testing brain tissue because of its fragile and tacky nature, and they are usually glued at the boundaries (brain/platen interface) as an alternative to clamping. This arrangement produces an inhomogeneous deformation field near the boundaries (see Miller and Chinzei [39]). These end effects contribute to higher magnitudes of stresses, thus resulting in steeper stress – strain curves. They also preclude the use of analytical tension – stretch relations.

Therefore, in this research, we focus on the development and calibration of a custom-designed High Rate Tension Device (HRTD) which is capable of performing tests at strain rates ≤ 90/s. Calibrations were performed with and without brain tissue specimens in order to ensure uniform velocity. In the second phase of this research, an appropriate specimen thickness and aspect ratio were determined in order to avoid any significant end effects due to inhomogeneous deformation of brain tissue during tensile tests. The inhomogeneous effects were investigated by performing several tensile tests with variable

sample thicknesses of 4.0, 7.0 and 10.0 mm while maintaining a constant nominal diameter of 15.0 mm at a strain rate of 30/s. The experimental data is also analyzed numerically as a linear elastic material using Young's modulus (E) as well as a nonlinear hyperelastic material by using the one-term Ogden model (μ, α) in the ABAQUS Finite Element code. This research will provide further insight into the behavior of brain tissue and the feasibility of performing reliable tension experiments on suitably sized specimens of brain tissue.

2 Materials and Method

2.1 Experimental Setup

A custom-designed *High Rate Tension Device* (HRTD) was designed and developed to perform tests at variable loading velocities (120 – 300 mm/s) to investigate inhomogeneous deformation effects on brain tissue at different specimen thicknesses, as shown in Fig.1. This apparatus is described here, with more comprehensive details being provided elsewhere [42]. The major components of the apparatus include a *servo motor controlled programmable electronic actuator* (700 mm stroke, 1500 mm/s velocity, *LEFB32T-700,* SMC Pneumatics), two ± *5* N *load cells* (rated output: 1.46 mV/V nominal, 150% safe overload of rated output, GSO series, Transducer Techniques) and a *Linear Variable Displacement Transducer* (range ± 25 mm, linearity ± 0.25 percent of full range output, ACT1000 LVDT, RDP Electronics). The load cells were calibrated against known masses and a multiplication factor of 13.67 N/V (determined through calibration) was used to convert voltage to load. An integrated single-supply instrumentation amplifier (AD 623 G = 100, Analog Devices) with a built-in single pole low-pass filter having a cut-off frequency of 10 kHz was used. The output of the amplifier was passed through a second single pole low-pass filter with a cut-off frequency of 16 kHz. The amplified signal was analyzed through a data acquisition system with a sampling frequency of 10 kHz. The force (N) and displacement (mm) data against time (s) were recorded for the tissue experiencing 30% strain. High speed image recording of brain tissue during tension tests was done at a frame rate of 3906 fps with 640 x 480 resolutions by using a high speed digital camera (Phantom V5.1, CMOS 10 bit Sensor, 1200 frames per second (fps) at maximum resolution - 1024 x 1024 and 95000 fps at a minimum resolution - 64 x 32). The images were examined to ensure that the faces of the specimens remained firmly bonded to the moving and stationary platens during extension of the brain specimen.

2.2 Calibration of Experimental Setup

Calibration of the HRTD was essential in order to ensure uniform velocity during extension of brain tissue at each strain rate. Two main contributing factors for the non-uniform velocity were the deceleration of the electronic actuator when it approached the end of the stroke and the opposing forces acting against the striking mechanism. Therefore, the striking mechanism (striker and shock absorber assembly as shown in Fig. 1) was designed and adjusted to ensure that it impacted the tension pin approximately 150 mm before the actuator came to a complete stop. The *striker* impact generated backward thrust, which was fully absorbed by the spring mounted on the actuator guide rod to prevent any damage to the *programmable servo motor*. Moreover, the actual actuator velocity was kept higher than the required (theoretically calculated) velocity to overcome the opposing forces acting against the striking mechanism (LVDT probe and sliding components). During the calibration process, the actuator was run several times with and without any brain tissue specimen to ensure repeatability of displacement (mm) against time (s). Fig 2 shows a typical output from the load cell and LVDT. After it was established that the actuator was capable of providing the required uniform velocity, the brain tissue specimen was mounted on the HRTD for the actual tests.

2.3 Specimen Preparation

Ten fresh porcine brains from approximately six month old pigs were collected from a local slaughter house and tested within 3 h postmortem. Each brain was preserved in a physiological saline solution at 4 to 5^{o}C during transportation. Then, 32 specimens were excised from 8 porcine brains (4 specimens from each brain). The dura and arachnoid were removed and the cerebral hemispheres were first split into right and left halves by cutting through the corpus callosum. As shown in Fig. 3, one half of the cerebral hemisphere was cut in the coronal plane to extract two coronal slices. Cylindrical specimens composed of mixed white and gray matter were prepared using a circular steel die cutter. The samples were then inserted in a cylindrical metal disk of 15.1 mm internal diameter to variable thicknesses of 4.0, 7.0 and 10.0 mm. The excessive brain portion was then removed with a surgical scalpel. The time elapsed between harvesting of the first and the last specimens from each brain was 14 ~ 17 minutes. Physiological saline solution was applied to the specimens frequently during cutting and before testing in order to prevent dehydration. The

specimens were not all excised simultaneously, rather each specimen was tested first and then another specimen was extracted from the cerebral hemisphere. All samples were prepared and tested at a nominal room temperature of 22 ^{o}C and relative humidity of 34 – 35%. Due to the extreme softness and tackiness of brain tissue, each specimen was tested only once and no preconditioning was done [23, 25, 39, 41].

2.4 Specimen Attachment Procedure

Reliably attaching soft tissue to the platens was very important in order to achieve high repeatability. To perform tests on the HRTD, the surfaces of the platens were first covered with a masking tape substrate to which a thin layer of surgical glue (Cyanoacrylate, Low-viscosity Z105880–1EA, Sigma-Aldrich) was applied. The prepared cylindrical specimen of tissue was then placed on the lower platen. The top platen, which was attached to the 5 N load cell, was than lowered slowly so as to just touch the top surface of the specimen. During the tests, the top platen remains stationary while the lower platen moves down to produce the required tension in the specimen as shown in Fig 1. One minute settling time was sufficient to ensure proper adhesion of the specimen to the platens.

Calibrating metal disks of 4.0, 7.0 and 10.0 mm thicknesses were also used to confirm the required distance between the platens before the start of experimentation. During tensile tests, excellent bonding was achieved at the brain/ platen interface, however inhomogeneous deformation of the brain tissue was observed at the edges of the brain/platen interface, as shown in Fig. 4. A high speed camera was used to monitor and record all tension tests and thus to confirm proper adhesion of brain tissue, as discussed in Section 2.2.

3 Hyperelastic Constitutive Modelling

3.1 Preliminaries

In general, an isotropic hyperelastic incompressible material is characterized by a strain-energy density function W which is a function of two principal strain invariants only: $W = W(I_1, I_2)$, where I_1 and I_2 are defined as [43],

$$I_1 = \lambda_1^2 + \lambda_2^2 + \lambda_3^2, \tag{1}$$

$$I_2 = \lambda_1^2\lambda_2^2 + \lambda_1^2\lambda_3^2 + \lambda_2^2\lambda_3^2. \tag{2}$$

Here $\lambda_1^2, \lambda_2^2, \lambda_3^2$ are the squares of the principal stretch ratios, linked by the relationship $\lambda_1 \lambda_2 \lambda_3 = 1$, due to incompressibility.

It was not possible to achieve homogeneous deformation conditions during tension tests due to the bonding of brain tissue (no slip conditions) at the platen/brain interfaces. This was considered to be a practical limitation of our experimental protocol. Nevertheless, an effort was made to select an appropriate specimen aspect ratio, S (diameter/thickness) which would produce negligible inhomogeneous deformation effects due to the no slip boundary conditions. Due to symmetry and incompressibility, the stretch ratios are of the form

$$\lambda_1 = \lambda, \quad \lambda_2 = \lambda_3 = \frac{1}{\sqrt{\lambda}} \tag{3}$$

where $\lambda \geq 1$ is the stretch ratio in the direction of tension. Also, Eqs. (1) and (2) give

$$I_1 = \lambda^2 + 2\lambda^{-1}, \; I_2 = \lambda^{-2} + 2\lambda, \tag{4}$$

so that W is now a function of λ only. During the experimental tension tests, the principal stretch ratio λ was calculated from the measure of the elongation e using equation: $\lambda = 1 + e$. The nominal stress component along the direction of tension S was evaluated as $S \equiv F / A$, where F is the tension force, as measured in Newtons by the load cell, and A is the area of a cross section of the sample in its undeformed state. The experimentally measured nominal stress was then compared to the predictions of the hyperelastic models from the following relation [43], valid for homogeneous tensile tests

$$S_{33} = \frac{\mathrm{d}\tilde{W}}{\mathrm{d}\lambda}, \text{ where } \tilde{W}(\lambda) \equiv W(\lambda^2 + 2\lambda^{-1}, \lambda^{-2} + 2\lambda), \tag{5}$$

3.2 Ogden Strain Energy Function

The Ogden model [44] has been used in the past to describe the nonlinear mechanical behavior of the brain, as well as of other nonlinear soft tissues [18, 39, 42, 45, 46]. Soft biological tissue is often modeled well by the Ogden formulation and most of the mechanical test data available for brain tissue in the literature are fitted with an Ogden hyperelastic function. The one-term Ogden hyperelastic function is given by

$$W = \frac{2\mu}{\alpha^2}\left(\lambda_1^\alpha + \lambda_2^\alpha + \lambda_3^\alpha - 3\right), \tag{6}$$

where $\mu > 0$ is the infinitesimal shear modulus, and α is a stiffening parameter. It yields the following nominal stress S_{33}, in the case of a homogeneous tensile test,

$$S_{33} = \frac{2\mu}{\alpha}\left\{\lambda^{\alpha-1} - \lambda^{-\left(\frac{\alpha}{2}+1\right)}\right\}. \tag{7}$$

4 Results

4.1 Experimentation

Ten tensile tests were performed at each specimen thickness of 4.0 ± 0.1 mm, 7.0 ± 0.1 mm and 10.0 ± 0.1 mm at a constant strain rate of 30/s up to 30% strain; however the diameter of each specimen was always 15.0 ± 0.1 mm. When measuring the dimensions of the specimens, it was noted that the nominal dimensions were reached after a few minutes; it was at this stage that testing commenced. All tests were conducted at a nominal room temperature of 22 ^{o}C. Each specimen was tested once and then discarded because of the highly dissipative nature of brain tissue. In order to maintain a constant strain rate with variable specimen thickness, the machine velocity was varied at each thickness. The required velocities for each specimen thickness and the actual measured velocities during each test are shown in Table 1.

Force (N) and displacement (mm) data were measured directly through the data acquisition system (Handyscope, HS4) at a sampling frequency of 10 kHz; this was converted to engineering stress (kPa) – time (s) for each specimen thickness. The experimental stress profiles obtained after experimentation are shown in Fig. 5. It is interesting to note that the stresses are significantly higher for the thinner specimens than for the thicker specimens at the same strain rate (30/s). It is also observed statistically, using a one-way ANOVA test, that there is a significant difference ($p = 0.000245$) between 4.0 and 7.0 mm specimen thicknesses and similarly ($p = 0.1614$) between 7.0 and 10.0 mm specimen thicknesses. This is due to the specimens being restricted at the boundaries (brain/platen interface) because they are attached to the platens using surgical glue. Because of this restriction, the tissue deformation near the platen ends is *inhomogeneous*. These end effects contribute to higher magnitudes of stress, thus resulting in steeper stress – strain curves. The maximum engineering stresses at the nominal specimen thicknesses of 4.0, 7.0 and 10.0 mm are 4.5 ± 1.244 kPa, 2.73 ± 0.44 kPa, 2.24 ± 0.75 kPa (mean ± SD), respectively. Numerical analysis using ABAQUS 6.9/Explicit was also carried out in order

to investigate further the effects of variable specimen thickness (4.0, 7.0, 10.0, 11.0, 12.0, 13.0, 14.0 mm).

4.2 Finite Element Simulation

Numerical simulations were performed on different specimen thicknesses by applying various boundary conditions using ABAQUS 6.9/Explicit to mimic experimental conditions. One end of the cylindrical specimen was constrained in all the directions whereas the other end was stretched up to 30% strain. Mass density $1040\,kg/m^3$ and C3D8R elements (hexagonal, 8-node linear brick with reduced integration) were used in the simulations. One-term Ogden material parameters (μ = 4395.0 Pa, α = -2.8) were obtained by performing inverse finite element analysis. The derived material parameters converged to average experimental engineering stress (kPa) – engineering strain data obtained at different specimen thicknesses (4.0, 7.0 and 10.0 mm). This procedure was particularly important to ensure that the resulting material parameters were the same irrespective of specimen thickness. The same procedure was adopted to determine Young's modulus E = 11200 Pa with a Poisson's ratio of 0.4999 as a linear elastic model, although of course the range of applicability of this model is much smaller than that of the non-linear Ogden model. The computed modulus was approximately equal to three times μ ($E = 3\mu$) as expected in incompressible elasticity. The derived elastic and hyperelastic parameters were kept constant for all the simulations performed at various specimen thicknesses.

A mesh convergence analysis was carried out by varying mesh density. In our study, we regarded the mesh as being convergent when there was a negligible change in the numerical solution (0.8%) with further mesh refinement. We achieved convergence with 2016 to 3598 elements for the 4 to 14.0 mm thick specimens of diameter 15.0 mm, respectively. The average simulation time for these models was 60s. We also analyzed the *accumulated artificial strain energy* used to control *hourglass deformation* during numerical simulations. It was observed that the *artificial strain energy* for the whole model, as a percentage of the *total strain energy*, was within the range of 1.66 – 3.4%. The significance of this low proportion of artificial strain energy (≤ 3.4%) indicates that hourglassing is not a problem.

Excellent agreement between the numerical engineering stress (using linear elastic and hyperelastic parameters) and the average experimental engineering stress (kPa) profiles is achieved, as shown in Fig. 6. It is established numerically and experimentally that the magnitudes of stresses are significantly higher with the reduction in specimen thickness or

at higher aspect ratio, S (diameter/thickness). The directly measured force (N) during the experimentation was also compared with the forces determined numerically. Excellent agreement is also achieved between the force – engineering strain profiles at different specimen thicknesses, as shown in Fig. 6. The stress behavior is further analyzed numerically at higher specimen thicknesses (11.0 to 14.0 mm) in order to confirm an appropriate aspect ratio which produces insignificant effects due to inhomogeneous deformation at the brain/platen interface. The numerically determined profiles of engineering stress at different sample thicknesses (4 – 14.0 mm) are shown in Fig. 7. Based on a one-way ANOVA statistical analysis, it is found that there is no significant difference ($p = 0.9196$) in the stress magnitudes between the thickest specimens as (10.0 – 14.0 mm) i.e., at low aspect ratios (S = 1.5 – 1.07). However, there is a statistically significant difference ($p = 0.000264$) between specimens of 4.0 and 7.0 mm thickness and similarly ($p = 0.12558$) between 7.0 and 10.0 mm thick specimens. The distribution of stresses (stress S33) and strains (true strain LE) were determined numerically using hyperelastic parameters (μ = 4395.0 Pa, α = -2.8) at variable specimen thicknesses.

It is clearly observed that a more homogeneous stress pattern is achieved at an aspect ratio of S (diameter/thickness) $\leq$ 1.5, however stresses are significantly higher in the case of the 4.0 mm thick specimen, as depicted in Fig. 8. Similarly, the strain distributions are also analyzed at different specimen thicknesses, as shown in Fig. 9. More homogeneous strain behavior is observed at aspect ratios S $\leq$ 1.5: this is clearly depicted in Fig. 9. The inhomogeneous strain effects at the platen ends are significantly reduced with the increase in specimen thickness. Similar stress and strain contours were also obtained when simulations were performed using a linear elastic model (Young's modulus, E = 11200 Pa) and Poisson's ratio of 0.4999.

4.3 Aspect Ratio Analysis

The diameter of a test sample is also an important factor to be considered when using cylindrical specimens. Therefore, numerical simulations were performed at 10.0, 15.0 and 20.0 mm diameters for each specimen thickness (4.0, 7.0, 10.0 and 13.0 mm). Stiffening behavior is observed with larger specimen diameters; however this effect is significantly reduced at the larger specimen thickness of 13.0 mm, as shown in Fig. 10. The larger diameter produces more inhomogeneous deformation which contributes to the higher magnitudes of stress. The difference between the stress profiles at aspect ratios S = 10/10

and 10/13 was also analyzed statistically using a one-way ANOVA test (p = 0 .1089). The stress magnitudes are slightly higher (7%) in case of S = 15/10 as compared to S = 10/10.

5 Discussion

Tensile tests on cylindrical specimens of brain tissue cannot be fully characterized as classical uniaxial tension because of the specimen restriction at the boundaries (brain/platen interface). There is a strongly inhomogeneous deformation field of the brain tissue near the boundaries because of its fixed attachment to the platens using surgical glue. The end effects contribute to higher magnitudes of stress, thus resulting in steeper stress – strain curves. Therefore, it was essential to determine an appropriate specimen thickness to avoid any significant end effects.

The effects of inhomogeneous deformation of brain tissue at the brain/platen interface have been analyzed experimentally and numerically using ABAQUS 6.9/Explicit. Experiments were performed using 4.0, 7.0, 10.0 mm specimen thicknesses while maintaining a constant nominal diameter of 15.0 mm. Excellent agreement was achieved between the average experimental engineering stress and the numerical engineering stress, as shown in Fig. 6. It was observed that the tensile stresses of the brain tissue are significantly different at variable specimen thicknesses. The analysis was extended further by numerical simulation of specimen thicknesses from 11.0 to 14.0 mm and at variable aspect ratios, S (diameter/thickness).

Based on the present analysis, it was determined that cylindrical specimens of aspect ratio S = 10/10 or lower (10/12, 10/13) are suitable to perform tensile tests on brain tissue. Larger aspect ratio specimens do not have a sufficiently uniform stress distribution to provide meaningful results. It is noted that Miller and Chinzei [39] used cylindrical samples of diameter 30.0 mm and height 10.0 mm (S = 3) during tensile tests at quasi-static velocities (0.005, 5.0 and 500 mm/min), whereas in compression tests they used a sample height of 13 mm (S = 2.3). Tamura et al [40], on the other hand, performed tensile tests at 0.9, 4.3 and 25/s strain rates using cylindrical specimens of diameter ~ 14.0 mm and height ~ 14.0 mm (S = 1.0).

During numerical simulations up to 30% strain, it was observed that one-term Ogden hyperelastic parameters (μ = 4395.0 Pa, α = -2.8) and linear elasticity (Young's modulus E = 11200 Pa) produced results which were in good agreement with experimental engineering stresses, as shown clearly in Fig. 6. The value of Young's modulus E = 11200 Pa is very

similar to that assumed by Morrison et al [7] (E=10 kPa) in finite element simulations to predict strain fields in a stretched culture of rat brain tissue. The results of the present study at a strain rate of 30/s are slightly lower than those measured by Tamura et al [40] who performed tests at a strain rate of 25/s (E = 18.6 ± 3.6 kPa).

The experimental results of this study are based on specimen dimensions of 15.0 mm diameter and variable sample thickness (4.0. 7.0 and 10.0 mm); thus there is a possibility of slightly higher stress magnitudes. Nevertheless, the primary objective of this research was to ascertain suitable specimen dimensions which should have minimum inhomogeneous deformation effects during tension tests. By following the procedure adopted in this study, the inhomogeneous deformation of the brain tissue can also be analyzed at a dynamic strain rate of 90/s in order to further understand the behavior of brain tissue at dynamic strain rates.

6. Conclusions

There are four important conclusions from this work.

(i) We have demonstrated the development and calibration of a custom designed HRTD that is useful to obtain experimental data up to moderate strain rates of 30/s. However, tests can be performed up to strain rates of 90/s using the same experimental setup.

(ii) We found that a brain specimen aspect ratio S = 10/10 or lower (10/12, 10/13) is most suitable for the tensile tests.

(iii) We estimated the one-term Ogden material parameters (μ = 4395.0 Pa, α = -2.8) which will prove useful for the nonlinear hyperelastic analysis of porcine brain tissue at a strain rate ~ 30/s.

(iv) We found the Young modulus E = 11200 Pa of the material in order to analyze the behavior of brain tissue in the small strain range.

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Figure Captions

Fig. 1 Schematic diagram of the complete test apparatus. Dashed and solid lines indicate inputs and outputs respectively from the electronic components.

Fig. 2 A typical displacement (mm) and force (N) signal against time (s) from data acquisition system at 10 kHz sampling rate.

Fig. 3. Extraction of cylindrical specimens from porcine brain tissue containing mixed white and gray matter

Fig. 4 – Experimentation using HRTD, which shows a cylindrical specimen (~ 15.0 mm diameter and ~ 10.0 mm thick) stretched to achieve 30% strain. Note the inhomogeneous deformation of tissue at the edges of the brain/platen interface.

Fig – 5. Stress profiles for different thickness specimens at a constant strain rate of 30/s up to 30% strain. Note that the increasing stress with thinner specimens is a consequence of inhomogeneous deformations at the specimen – platen interfaces.

Fig. 6 – Good agreement of elastic, hyperelastic and experimental stress (kPa) and force (N) profiles at different sample thicknesses, using Ogden material parameters (μ = 4395.0 Pa, α = -2.8) and Young's modulus E = 11200 Pa (at 10% strain).

Fig. 7 – Decease in the magnitude of engineering stress with increasing specimen thickness, as determined numerically from the material parameters (μ = 4395.0 Pa, α = -2.8)

Fig. 8 Stress contours at variable specimen thicknesses, at a maximum stretch ratio of 1.3. Significantly higher stresses are evident for specimens of 4.0 mm thickness. S = aspect ratio of diameter/thickness.

Fig – 9. Strain contours at variable thickness, at a maximum stretch ratio of 1.3. Homogeneous strain field is evident at 10.0 mm specimen thickness and above i.e., $S \leq 1.5$.

Fig – 10. Variation in engineering stress profiles at different aspect ratios, S = diameter/thickness.