Title	Mechanical characterisation of polyurethane elastomer for biomedical applications
Authors(s)	Kanyanta, Valentine, Ivankovic, Alojz
Publication date	2010-01
Publication information	Kanyanta, Valentine, and Alojz Ivankovic. "Mechanical Characterisation of Polyurethane Elastomer for Biomedical Applications" 3, no. 1 (January, 2010).
Publisher	Elsevier
Item record/more information	http://hdl.handle.net/10197/4615
Publisher's statement	This is the author's version of a work that was accepted for publication in Journal of the Mechanical Behavior of Biomedical Materials. Changes resulting from the publishing process, such as peer review, editing, corrections, structural formatting, and other quality control mechanisms may not be reflected in this document. Changes may have been made to this work since it was submitted for publication. A definitive version was subsequently published in Journal of the Mechanical Behavior of Biomedical Materials (3, 1, (2010)) DOI: http://dx.doi.org/10.1016/j.jmbbm.2009.03.005
Publisher's version (DOI)	10.1016/j.jmbbm.2009.03.005

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Mechanical Characterisation of Polyurethane Elastomer for Biomedical Applications

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1. Introduction

Polyurethane elastomers have been widely used as biomaterials, with applications ranging from medical devices and utilities like cardiac-assist pumps and blood bags, to chronic implants such as heart valves and vascular grafts (Gogolewski, 1992; Christenson et al., 2005; Lelah et al., 1986; Szycher et al., 1992; Stokes et al., 1995; Gunatillake et al., 2003). They were first proposed for use as biomaterials in 1967 by Boretos and Pierce (1967). Their superior mechanical properties and blood compatibility has favored their use and development as biomaterials, particularly as components of implanted devices (Lamba et al., 1997). Polyurethane elastomers offer superior mechanical properties over silicone elastomers, particularly in relation to tear and abrasion, and flex-fatigue life (Wiggins et al., 2003). The chemical composition of these elastomers offers substantial opportunities to synthetic polymer chemists to tailor the structures to meet specific requirements. Generally polyurethanes offer good compatibility with tissue and blood, with a good resistance to mechanical degradation. Thermoplastic polyurethane elastomers are a class of linear segmented copolymers characterised by the presence of urethane (carbamate) groups. They are prepared from three components: a diisocyanate, a macrodiol, and a chain extender (Gunatillake et al., 2003), and can be categorised into two major groups depending on the macrodiol used i.e. ester based or ether based polyurethane (Knoerr and Hoffmann; Recker, 2001). Much research and developments have been done in achieving polyurethanes with excellent biostability with a combination of good mechanical properties and biocompatibility. Polyurethanes used in chronic implants are subject to hostile *in vivo* conditions which can lead to their failure (Stokes et al., 1995; Wiggins et al., 2003;

Thoma, 1987; Schmidt et al., 1998; Zhao et al., 1990, 1991; Wu et al., 1992; Schubert et al., 1995, 1997).

In the current work, an ether based polyurethane elastomer supplied by Renier Technology Ltd (UK) is used for mock arteries, representing large arteries. Like most biological materials, the arterial wall has a very complex mechanical behaviour. It exhibits strong mechanical anisotropy, nonlinear stress-strain behaviour, viscoelasticity and poroelasticity (Julia et al., 1966; L'Italien et al., 1994). Its stress-strain behaviour is also dependent on local hemodynamics and shear stress (Stephanie et al., 1998; Glagov, 1994; Barbee et al., 1994; Tronc et al., 1996), age and inactivity (Vaitkevicious et al., 1993; Kelly et al., 1989; Avolio et al., 1985), pathological infections (David et al., 2002) and high blood pressure (Avolio et al., 1985), and vary widely along the arterial tree (Mangel et al., 1996). The arterial wall is composed of the elastin, vascular smooth muscle cells (VSMC) and connective tissue (Figure 1). Inside, the wall is covered by a monolayer of endothelial cells and epithelial cells form its outside. Due to the complex structure and behavior, it is extremely difficult to find a material with similar mechanical behaviour to arteries or to model it by a single hyperelastic material model. Although the properties of polyurethane elastomer differ from those of the arteries, it was chosen because of its good biocompatibility and viscoelastic properties, and for its wide use in vascular grafts (Szycher, 1998; Gunatillake et al., 2003), making it a preferred material for mock arteries in this work.

Since constitutive material models can not entirely describe the stress-strain behaviour of the material under all loading conditions due to enormous variations and complexity involved, material tests were designed based on loading conditions representative of the application of interest. These conditions were chosen to be representative of *in vivo* loading conditions of large arteries. Material tests included uniaxial tests under varying temperature, humidity and strain rates, planar and equibiaxial tension, relaxation, creep and cyclic tests on specimens made of polyurethane elastomer. Cyclic tests were performed to establish viscoelastic properties of the elastomer relevant for

pulsatile pressure loading of mock arteries. Also performed were static and dynamic flow tests on polyurethane tubular specimens representing mock arteries, which are discussed in detail in (Quinn et al., 2008).

Although polyurethane elastomer is widely used, its mechanical behaviour has never been extensively characterised. The material properties of the elastomer are usually measured in uniaxial tension only, under dry room temperature and at low strain rates (Diaconu et al., 2006; Pathiraja et al., 1996; Diaconu et al., 2005). This work establishes the variations in the behaviour of polyurethane elastomer with temperature, humidity and strain rate and also reports planar and equibiaxial tension, relaxation, creep and cyclic test results, providing a comprehensive characterisation of polyurethane elastomer.

This work is part of the study looking at the role of hemodynamic shear stress in atherosclerosis, by predicting accurately the distribution and rate of change of wall shear stress in various arterial geometries (straight and branched mock arteries) and establishing the role of arterial flexibility on wall shear stress.

2. Test Methods

Material tests performed on polyurethane elastomer specimens included uniaxial, planar and equibiaxial tension, stress relaxation, creep and cyclic loading. The test specimens were of different geometries (Figure 2), according to the test being performed. All the stresses and strains presented in the plots in this work refer to the engineering stress and strain, respectively.

2.1. Uniaxial Tension

Uniaxial tension tests were performed using dumbbell shaped specimens (Figure 2(a)). These tests can be divided into three groups, according to the strain rates employed i.e. low ($< 1/s$), intermediate (between $1/s$ and $100/s$) and high ($> 100/s$) strain rate uniaxial tension tests.

Low strain rate tests were performed under three different conditions; dry-room temperature, wet-room temperature and wet@37°C, on a standard Instron machine. A laser extensometer was used for a non contact measurement of strains. This is important when measuring strains of soft materials as any strain measurement involving contact could affect the stiffness of the specimen. An environmental chamber, equipped with a thermostat controlled heater, was used to maintain the test temperature constant (e.g. 37 °C for wet@37 °C condition) for the duration of the test. Specimens were submerged in water at the set temperature for one hour prior to testing, and remained submerged for the duration of the test.

Intermediate and high strain rate uniaxial tests were performed under dry-room temperature condition only, due to the current experimental limitations. An instrumented drop-weight tester (Figure 3) and split Hopkinson pressure bar in tension (Figure 4) were used for intermediate and high strain rate uniaxial tests, respectively.

For intermediate strain rate tests, load measurements were achieved with a calibrated strain gage, while specimen strains were obtained from the displacement of the striker (Figure 3), with the zero position defined as the point of initial contact between the striker and the impact plate.

Under high strain rate tests (Figure 4), a swinging pendulum was used to load the specimens. When the pendulum striker hits the block, a tensile stress pulse is generated in the incident bar. A dumbbell shaped cylindrical fracture piece made of Perspex (2.5 mm in diameter) breaks on impact, thus, preventing multiple pulses being generated. The amplitude of the generated pulse is equal to stress required to break the fracture piece, which in this case corresponds to a stress of 1.4 MPa and 57 MPa in the incident bar and fracture piece, respectively. In this way the amplitude of the incident pulse can be controlled by varying the diameter of the fracture piece (Shim et al., 2001). The generated tensile pulse propagates through the incident bar into the specimen. At the bar-specimen interface, part of this pulse is transmitted into the specimen and propagates through the transmitter

bar as a tensile pulse. The rest of the pulse reflects into the incident bar as a compressive pulse. The transmitted and reflected pulses are recorded by the transmitter and incident strain gages, respectively.

Specimen stress σ and strain ϵ are obtained from the transmitted and reflected strain signals, respectively, using the classical Kolsky analysis given by

$$\sigma = \frac{A_b}{A_s} \left(\frac{1 + R}{1 + R_b} \right) \epsilon_t \quad (1a)$$

$$\epsilon = \frac{A_b}{A_s} \left(\frac{1 - R}{1 + R_b} \right) \epsilon_t \quad (1b)$$

where A_b and A_s are the bar and specimen cross-section areas respectively, ϵ_t is the transmitted strain signal, ϵ_r is the reflected strain signal, l_s is the specimen gage length and C_b is the wave speed through the bar (i.e. $C_b = \sqrt{E/\rho}$). The material used for the incident and transmitter bars was a Glass Filled Nylon 66 with a Young's modulus $E = 13.2 GPa$, density $\rho = 1460 \text{ Kg/m}^3$, and diameter $D = 16 \text{ mm}$.

Ten specimens were tested at each strain rate for low strain rate uniaxial tests and five specimens at each strain rate for intermediate and high strain rate uniaxial tests.

2.2. Choice of Maximum Loading Rate

A maximum strain rate for the tests was chosen as the upper limit of the range of strain rates experienced by arteries *in vivo*. Strain rates in arteries can be estimated analytically, numerically or experimentally if the pressure waveform and arterial wall properties are known. The average Young's modulus for the arterial wall is reported to range between 600kPa and 1MPa (Riley et al., 1992). Taking for instance the pressure waveform for the carotid artery (Figure 5(a)) (Augst et al., 2007), strain rates, $\dot{\epsilon}_h = d(pd/2Eb)/dt$, in the artery can be estimated as shown in Figure 5(b) (assuming the artery's internal diameter $d = 10 \text{ mm}$, thickness $b = 1 \text{ mm}$ and Young's modulus $E =$

800kPa. p is the pressure). Therefore, the maximum strain rates in arteries can be assumed to be about 1/s. However, strain rates of up to 175/s were used in the design of uniaxial tension tests in order to fully establish the strain rate behaviour of polyurethane rubber, although this may not be particularly necessary for the current study.

2.3. Uniaxial Stress relaxation and Creep

Creep tests were performed by holding a specimen at a constant tensile stress and measuring the resulting strain ϵ as a function of time. Similarly, relaxation tests were performed by holding a specimen at a constant tensile strain and measuring the resulting stress σ as a function of time. Stress (for creep tests) and strain (for relaxation tests) were automatically controlled by use of QMAT Materials Testing and Analysis Software.

Two stress magnitudes, 0.44 MPa and 1.136 MPa, were used for creep tests and four strain magnitudes, 5%, 6.5%, 8% and 10%, for stress relaxation tests. Five specimens were tested at each stress or strain magnitude. Material stress relaxation and creep were allowed for 2000s in each case.

2.4. Uniaxial Cyclic Tests

Under uniaxial cyclic loading, the loading of the specimens was periodically repeated over 10 cycles. The loading function was either force (stress) or displacement (strain), which repeatedly took the form of equation (2). Constant triangular stress or strain cycles were applied, while the other quantity (strain or stress) is allowed to vary according to material behaviour (Sandor, 1972). The experimental set-up is similar to uniaxial tension tests and dumbbell shaped specimens (Figure 2(a)) were used. Four specimens were tested with either stress or strain as a loading function, making a total of eight specimens. Tests were performed at a loading rate of 10mm/min (corresponding to a strain rate of 0.013/s), under dry-room temperature condition.

$$\epsilon(t), \sigma(t) = a * \begin{cases} t; & 0 \leq t \leq T/2 \\ T-t & T/2 \leq t \leq T \end{cases}, \quad (2)$$

where “a” is the loading rate (strain or stress per unit time) and “T” is the period of the triangular function.

2.5. Planar tension

Specimens used for the planar tension tests were 6 times wider than their gage length (Figure 2(b)). This was in order to achieve a deformation, F , and stress, σ , state given by (assuming the elastomer behaves as an incompressible material)

$$\left[\text{Diagram 1} \right] \text{ and } \left[\text{Diagram 2} \right]. \quad (3)$$

The material is constrained in the transverse direction and only deforms in the loading and thickness directions (Miller, 1999). Apart from a different specimen type, the experimental set-up is the same as for simple tension. Tests were conducted under dry-room temperature condition at a strain rate of 0.013/s.

2.6. Equibiaxial tension

Cruciform type specimens (Figure 2(c)) were used for equibiaxial tension tests. The experimental set-up is as shown in Figure 6. The material was loaded in both directions, at a constant strain rate of 0.013/s, in such a way that the stretches in the two loading directions are identical i.e. $\lambda_1 = \lambda_2$. This was achieved by using a system of pulleys shown in Figure 6(b). The specimen deformation, F , and stress, σ , state are

$$\left[\text{Diagram 1} \right] \text{ and } \left[\text{Diagram 2} \right]. \quad (4)$$

Load measurements were achieved with the Load cell while specimen strains were obtained from the displacement of the cross head. Tests were conducted under dry-room temperature condition, and stress and strain were measured in one direction only.

2 mm diameter nylon strings with a Young's modulus of 3.9 GPa were used as connecting strings (Figure 6(b)). Considering that the maximum load for the tests is in the range of 20-30 N, deformations in the nylon strings is expected to be insignificant.

3. Hyperelastic Material Models

For isotropic hyperelastic materials, the strain-energy function W can be defined in terms of strain invariants $I_i (i = 1, 2, 3)$ or principal stretches $\lambda_i (i = 1, 2, 3)$ i.e.

$$W = W(I_1, I_2, I_3) \quad \text{or} \quad W = W(\lambda_1, \lambda_2, \lambda_3) \quad (5)$$

If incompressibility is assumed (i.e. $I_3 = 1$), equation (5) then reduces to a function of two variables only i.e. I_1 and I_2 . Therefore, for an isotropic, incompressible material, the principal Cauchy stresses in the loading axis for uniaxial, planar and equibiaxial tension are given by equations 6, 7 and 8, respectively (Holzapfel, 2000).

$$\sigma_{11} = 2 \left(\lambda^2 - \frac{1}{\lambda} \right) \left[\frac{\partial W}{\partial I_1} + \frac{1}{\lambda} \frac{\partial W}{\partial I_2} \right]. \quad (6)$$

$$\sigma_{11} = 2 \left(\lambda^2 - \frac{1}{\lambda^2} \right) \left[\frac{\partial W}{\partial I_1} + \frac{\partial W}{\partial I_2} \right]. \quad (7)$$

$$\sigma_{11} = \sigma_{22} = 2 \left(\lambda^2 - \frac{1}{\lambda^4} \right) \left[\frac{\partial W}{\partial I_1} + \lambda^2 \frac{\partial W}{\partial I_2} \right]. \quad (8)$$

4. Results

The results of the material tests show the behaviour of the polyurethane elastomer tested to be highly dependent on temperature and humidity, as shown in Figure 7(a). The elastomer significantly

softens with increase in temperature and humidity levels, with the Young's modulus of 7.4 MPa, 5.3 MPa and 4.7 MPa for dry-room temperature, wet-room temperature and wet@37°C conditions, respectively. These results agree with the reported observations that polyurethane based medical invasive devices soften significantly within minutes of insertion into a human body, resulting in reduced patient discomfort and risk of vascular trauma (Tilak, 2001). The reported Young's modulus of polyurethane elastomer at dry room temperature varies widely depending on elastomer composition, with values of 3.6 MPa (Diaconu and Dorohoi, 2005), 13.1 MPa (Diaconu et al., 2006), and 14.6 – 88.8 MPa (Pathiraja et al., 1996). All the stresses and strains presented in the plots in this paper refer to the engineering stress and strain, respectively.

Figure 7(b) shows the stress-strain behaviour of polyurethane elastomer under uniaxial, planar and equibiaxial tension, for a strain rate of 0.013/s and under dry-room temperature condition. At low strain rates ($< 1/s$), the elastomer shows little or no strain rate dependency (Figure 8). Therefore, under this range of strain rates, the behaviour of polyurethane elastomer can be considered strain rate independent.

Intermediate strain rate uniaxial tension test results are presented in Figure 9, for three different strain rates. Under this range of strain rates, polyurethane elastomer shows considerable strain rate sensitivity, with the Young's modulus of 8 MPa, 9.5 MPa and 10.5 MPa for the strain rates of 29.4 /s, 58.8 /s and 88.2 /s, respectively. There is significant scatter in the results of the tests at 88.2 /s, above 30% strains (Figure 9(c)). This is probably due to the inaccuracy in conducting these tests as the impact speed increase. However, for strains below 20%, the results are very repeatable for all the three strain rates.

Figure 10 presents high strain rate test results for a strain rate of 137.5 /s. The incident and reflected strain signals, recorded by the incident strain gage, are shown in Figure 10(a), while the transmitted strain signals (recorded by the transmitted strain gage) are presented in Figure 10(b). The noise in the transmitted signal is significant because of the high amplification used, owing to the

small signal amplitude. The averaged results are shown in Figure 10(c), and the reflected and transmitted strain signals, which are aligned to have a common starting point in time, in Figure 10(d).

Applying Kolsky analysis (equation (1)) on the transmitted and reflected strain signals yields the stress-strain curves shown in Figure 11, showing the loading phase and part of the unloading phase, for the strain rates of 137.5 /s and 175 /s. The unloading curve does not follow the same path as the loading curve, probably due to the interference in the transmitted signal (Figure 10(c)) caused by the re-reflected waves (resulting in 2nd stage loading of the specimen). The material's Young's modulus at these strain rates is 12 MPa and 13.5 MPa, for 137.5 /s and 175 /s strain rates, respectively. Figure 12 presents the change in the stiffness of polyurethane elastomer with increase in strain rate. The Young's modulus linearly increases with the strain rate at a rate of 0.034 MPa per strain rate.

Viscoelastic properties of polyurethane elastomer were determined from stress relaxation and creep tests by fitting a 3-term Prony series to the relaxation and creep data. Figure 13 presents the relaxation and creep data, and the calculated relaxation and creep compliance for different strain and stress magnitudes, respectively. At 2000s, the material can be considered stabilized (when the material modulus reaches a steady-state value) since the difference in the stress at 500s and 2000s is less than 2 % (Figure 14 (a)). The material's relaxed modulus is determined by plotting the relaxed data at 2000s against the strains (Figure 13(e)). This value is 6.5 MPa, and is 88 % of the initial or unrelaxed modulus. The relaxation modulus $E(t)$ and creep compliance $J(t)$ functions, determined by fitting the Prony series to the relaxation data (Figure 14(a)) and creep data (Figure 14(b)), respectively, are given by

$$\boxed{\phantom{E(t) = E_{\infty} \left(1 - \sum_{i=1}^N g_i \left(1 - \exp\left(-\frac{t}{\tau_i}\right) \right) \right)}} \quad (9a)$$

$$\boxed{\phantom{J(t) = J_{\infty} \left(1 - \sum_{i=1}^N g_i \left(1 - \exp\left(-\frac{t}{\tau_i}\right) \right) \right)}} \quad (9b)$$

The averaged uniaxial cyclic test results (for eight specimens) are shown in Figure 15, with stress as a loading function (Figure 15(a)) and strain as a loading function (Figure 15(b)). The results show that the behaviour of polyurethane elastomer under cyclic loading exhibits creep and stress relaxation behaviours similar to those observed under static conditions or monotonic loading (Figure 13). When stress is applied as a loading function, the material undergoes creep as it deforms, which follow the same pattern as the creep data at 0.44 MPa constant stress (Figure 15(a)). Similarly, when strain is used as a loading function, the material undergoes stress relaxation (Figure 15(b)). Stress relaxation and creep behaviours in cyclic loading can be described in the same manner as under static conditions. After about 500s (or 10 loading cycles), there is little or no significant changes in the peaks of either stress or strain with successive loading cycles (maximum changes are below 2 % at this point). The material can hence be deemed stabilised after this point.

4.1. Hyperelastic Material Models

In this work, the polyurethane elastomer tested is treated as an isotropic, incompressible material. These assumptions are valid within experimental reason since the material anisotropy is only 10 % and the Poisson ratio ν is 0.475 +/- 0.025 (Kanyanta, 2008). Therefore, isotropic, incompressible hyperelastic material models were used in the choice of a representative material model for the elastomer

Six hyperelastic material models were fitted to the uniaxial, planar and equibiaxial tension experimental data using equations 6, 7 and 8, respectively, as shown in Figure 16. The strain energy functions for each material model are given in the appendix.

In order to relate the Cauchy stress in equations 6, 7 and 8 to the engineering stress σ_{eng} in experimental results, equation (13) was used.

$$\sigma_{11} = \sigma_{eng} \frac{A}{A_f} = \sigma_{eng} \lambda, \quad (13)$$

where A and A_f are the original and deformed cross-section areas respectively.

Yeoh's model provides the best fit to the range of experimental data (Figure 17(e)), with a R-squared value of 0.996. However, other models such as Neo-Hookean model, can be sufficient if one is only concerned with small strain deformations (below 15 %)

5. Discussion

The effects of humidity and temperature on material properties of polyurethane elastomers are usually not reported. This study presents a comprehensive characterisation of an ether-based polyurethane elastomer and reports the variations in its properties with humidity, temperature and strain rate. The behaviour of the elastomer is highly dependent on humidity, temperature and strain rate, with the Young's modulus of 7.4 MPa, 5.3 MPa and 4.7 MPa, at dry-room temperature, wet-room temperature and wet@37°C, respectively, and at strain rates below 1/s. Thus, the stiffness of polyurethane implants inside a human body is significantly different from that at dry-room temperature. This agrees with earlier observations that polyurethane elastomer products significantly soften on insertion into a human body (Tilak, 2001), thus, reducing a patient's post operation vascular trauma. A similar trend in humidity and temperature dependency behaviour is expected for all grades of polyurethane elastomers due to the strong similarities in their basic chemical structures (Knoerr and Hoffmann; Recker, 2001).

The Young's modulus of the elastomer also increased significantly at high strain rates with values of 7.4 MPa, 8 MPa, 9.5 MPa, 10.5 MPa, 12 MPa and 13.5 MPa, at strain rates of <1/s, 29.4/s, 58.8/s, 88.2/s, 137.5/s and 175/s, respectively, at dry-room temperature. However, strain rates in arteries are not expected to exceed 2/s (twice the estimated upper bound strain rate, Section 2.2). This also applies to the current study using polyurethane mock arteries. The Young's modulus only varies by about 0.5 % between 1/s and 2/s (Figure 12). Therefore, the Young's modulus at low strain rates (< 1/s) is sufficient to describe the stiffness of the elastomer for the range of strain rates in

arteries. Intermediate and high strain rates were only used to fully describe the strain rate dependent behaviour of polyurethane elastomer, but were not particularly necessary in this study.

Most literature values for the Young's modulus of polyurethane elastomer are measured at dry-room temperature and low strain rates. These include 3.6 MPa (Diaconu and Dorohoi, 2005), 13.1 MPa (Diaconu et al., 2006), and 14.6 – 88.8 MPa (Pathiraja et al., 1996). The wide variation in these values is due to the wide variety in the composition of polyurethane elastomers. These values are not adequate for modeling the behaviour of e.g. polyurethane implants inside a human body (at wet@37 deg Celsius) since the stiffness of polyurethane elastomer is seen to be highly dependent on humidity and temperature.

Polyurethane elastomer exhibits significant viscoelastic behaviour. Stress relaxation and creep behaviour under cyclic loading also compares well with that seen under monotonic loading. Most importantly is that, under cyclic loading, the elastomer reaches a steady-state behaviour within the first 10 loading cycles. This means that the material properties measured under monotonic loading can be used to model the behaviour of the elastomer under cyclic loading, assuming the material has reached steady-state conditions. The relaxed modulus at dry-room temperature was 14 % (6.5MPa) less than the initial modulus (7.4 MPa). Since creep and relaxation tests could not be conducted at wet-room temperature and wet-37deg Celsius, the same trend in material relaxation is assumed. Future work will clarify this. A 3-term Prony series was sufficient to model the elastomer's viscoelastic behaviour (Figure 14).

The elastomer also exhibits 10 % anisotropy and its Poisson ratio ranged between 0.45 - 0.5. In the current study, 10 % anisotropy is deemed insignificant and the Poisson ratio is treated to be close enough to 0.5. Therefore, the elastomer is assumed as an isotropic, incompressible material, and standard isotropic, incompressible hyperelastic material models are used in the choice of a representative material model for the elastomer. Yeoh's model was found to be the best representative material model for this elastomer. However, other models such as Neo-Hookean

model, which is much simply to implement in numerical codes, can be sufficient if one is only concerned with small strain deformations (below 15 %).

The chemical and bio-stability of the elastomer was not examined in this study. This is adequately covered in literature (Stokes et al., 1995; Wiggins et al., 2003; Thoma, 1987; Schmidt et al., 1998; Zhao et al., 1990, 1991; Wu et al., 1992; Schubert et al., 1995, 1997).

6. Conclusion

The behaviour of the ether-based polyurethane elastomer tested is humidity, temperature and strain rate dependent. Its Young's modulus is almost 40% less at wet@37 deg Celsius than it is at dry-room temperature, and linearly increases with increase in strain rate at a rate of 0.034 MPa/strain rate. For the range of strains rates found in arteries, i.e. 0 – 4 /s, there is insignificant variation in the Young's modulus of the elastomer and, thus, the Young's modulus can be assumed constant. The elastomer also exhibits 10 % anisotropy and 13 % viscoelasticity. The Yeoh model is the best representative material model for this elastomer, over a range of strains up to 300 %.

Although there is a wide variety of polyurethane elastomers used as biomaterials, the trend in their humidity and temperature dependency and cyclic and viscoelastic behaviour are expected to be very similar. Therefore, material properties measured at dry-room temperature can not be taken as the properties for polyurethanes used for chronic implants at wet@37 deg Celsius. Therefore, it is important that the properties of the elastomer are measured at conditions mimicking those of the intended application.

Acknowledgments

The authors wish to thank Science Foundation Ireland for the financial support of this work, and Dr. N. Murphy from UCD for constructive discussions and review of this manuscript.

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Appendix

Hyperelastic material models and their strain energy functions

NeoHookean:

$$W = \frac{\mu}{2} (I_1 - 3)$$

, μ is the shear modulus.

2-term Mooney Rivlin:

$$W = \frac{\mu_1}{2} (I_1 - 3) + \frac{\mu_2}{2} (I_2 - 3)^2$$

3-term Mooney Rivlin:

$$W = \frac{\mu_1}{2} (I_1 - 3) + \frac{\mu_2}{2} (I_2 - 3)^2 + \frac{\mu_3}{2} (I_3 - 3)^3$$

Arruda-Boyce:

$$W = \frac{\mu}{2} \left(I_1 - 3 + \frac{1}{\lambda - 1} \left(I_1^{\lambda} - 3^{\lambda} \right) \right)$$

where

$$I_1 = \text{tr}(\mathbf{F}^T \mathbf{F}), \quad I_2 = \frac{1}{2} \left((\text{tr}(\mathbf{F}^T \mathbf{F}))^2 - \text{tr}(\mathbf{F}^T \mathbf{F} \mathbf{F}^T \mathbf{F}) \right), \quad I_3 = \det(\mathbf{F})^2$$

Yeoh:

$$W = \sum_{i=1}^3 C_i (I_1 - 3)^i$$

Ogden:

$$W = \sum_{i=1}^n \frac{\mu_i}{\alpha_i} \left(\lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3 \right) \quad n = 3.$$

Figures

Figure 1: Arterial wall composition.

Figure 2: Specimen geometries for (a) uniaxial tension, cyclic, stress relaxation and creep, (b) planar tension and (c) biaxial tension tests. All dimensions are in mm

Figure 3: Experimental set-up for intermediate strain rate tests

Figure 4: Schematic of the high strain rate test on a Split Hopkinson pressure bar in tension.

Figure 5: Numerical estimation of strain rates in the carotid artery. (a) Pressure waveform and (c) hoop strain rates.

(a) (b)
Figure 6: Biaxial (a) experimental set-up and (b) schematic drawing of experimental set-up

(a) (b)
Figure 7: (a) Stress-strain behaviour of polyurethane rubber under different test conditions and (b) uniaxial, planar and equibiaxial tension stress-strain behaviour of the rubber under dry-room temperature condition, and at a strain rate of 0.013/s.

Figure 8: Strain rate dependency of polyurethane rubber at low strain rates.

(a)

(b)

(c)

(d)

Figure 9: Intermediate strain rate uniaxial test results for (a) 29.4 /s, (b) 58.8 /s, (c) 88.2 /s strain rates and (d) the average stress-strain behaviour of polyurethane rubber at different strain rates.

Figure 10: (a) Strain signals recorded by the incident and (b) transmitter strain gages, (c) the averaged strain signals and (d) reflected and transmitted strain signals, at a strain rate of 137.5 /s.

Figure 11: Stress-strain behaviour of polyurethane rubber at the strain rates of 137.5 /s and 175 /s.

Figure 12: The Young's modulus of polyurethane rubber expressed as a function of strain rate.

(a)

(b)

(c)

(d)

(e)

Figure 13: Viscoelastic behaviour of polyurethane rubber under dry-room temperature condition: (a) relaxation data and (b) relaxation modulus at different strain magnitudes, (c) creep data and (d) creep compliance at different stress magnitudes and (e) relaxed modulus.

(a)

(b)

Figure 14: Viscoelastic properties of polyurethane rubber: (a) Relaxation modulus, (b) Creep compliance

(a)

(b)

Figure 15: Cyclic tests results under dry-room temperature condition; (a) material creep, with stress as the loading function, and (b) material stress relaxation, with strain as a loading function

Figure 16: Comparison between experimental data and the predictions of hyperelastic material models for (a) NeoHookean, (b) 2-term Mooney Rivlin, (c) 3-term Mooney Rivlin, (d) Ogden, (e) Yeoh and (f) Arruda-Boyce material models.