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Development of novel synthetic muscle tissues for sports impact surrogates

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Title: Development of Novel Synthetic Muscle Tissues for Sports Impact Surrogates

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Abstract

Impact injuries are commonplace in sport and often lead to performance detriment and debilitation. Personal Protective Equipment (PPE) is prescribed as a mandatory requirement in most sports where these impacts are likely to occur, though the methods of governance and evaluation criteria often do not accurately represent sports specific injury scenarios. One of the key shortcomings of such safety test standards is the human surrogate to which the PPE is affixed; this typically embodies unrepresentative geometries, masses, stiffness and levels of constraint when compared to humans.

A key aspect of any human surrogate element is the simulant material used. Most previous sports specific surrogates tend to use off-the-shelf silicone blends to represent all the soft tissue structures within the human limb segment or organ; this approach potentially neglects important human response phenomena caused by the different tissue structures.

This study presents an investigation into the use of bespoke additive cure Polydimethylsiloxane (PDMS) silicone blends to match the reported mechanical properties of human relaxed and contracted skeletal muscle tissues. The silicone simulants have been tested in uniaxial compression through a range of strain rates and fit with a range of constitutive hyperelastic models (Mooney Rivlin, Ogden and Neo Hookean) and a viscoelastic Prony series.

Keywords

Polydimethylsiloxane, Silicone, Compression, Mechanical Testing, PPE, Simulant, Sports, Muscle

Highlights

- Bespoke PDMS silicones formulated to better match relaxed and contracted muscle.
- PDMS simulants tested in uniaxial compression through a range of strain rates.
- Constitutive material models have been evaluated and fitted to experimental data.

Abbreviations: PPE - Personal Protective Equipment; PDMS - Polydimethylsiloxane; FE - Finite Element; SHPB - Split Hopkinson Pressure Bar; **PVA - Polyvinyl Acetate**

1. Introduction

1.1. Background

In contact sports, personal protective equipment (PPE) is often a mandatory requirement to ensure the safety of the participant is maintained throughout play and to prevent injuries that would otherwise restrict the ability of players to perform at a maximal level. At a professional level these injuries can result in a financial burden to clubs and sponsors; at a recreational level these injuries can result in avoidable healthcare costs and a burden to society in terms of days absent from the workplace.

Sports PPE is governed by safety standards (e.g. BS:EN 13546:2002; BS:6183-3:2000; BS:EN 13061:2000) which prescribe the form and minimal acceptable levels of impact protection that the garment must embody (Fig. 1). The standards are predominantly concerned with risks associated with blunt trauma from compressive impacts and dictate the nature of the striker and surrogate bodies employed to simulate impact. These are typically high-mass, rigid anvil surrogates with unrepresentative target segment geometries rigidly constrained to a massive base. The nature of this surrogate and the likely contact interactions will generate impact phenomena unrepresentative of those experienced by the human body in sport. For example, the proportions of strain energy absorbed by the surrogate and PPE will differ significantly between a stiff and compliant surrogate.

Figure 1 – Schematic showing a typical sports safety standard protocol used to assess projectile impacts (BS:13546:2002 Field Hockey Goalkeeper Protective Clothing; BS:6183-3:2000 Cricket Leg Protectors)

Due to ethical restrictions, i.e. the risk of subject injury, human surrogates are critical in the development of new sports PPE during two phases of the development cycle. Firstly, research conducted by manufacturers to better understand the performance of current PPE requires surrogates with greater biofidelity to observe product interactions with soft tissues influencing performance failure but that are not replicated in standard tests. Performance prediction/analysis through computational simulation is a worthwhile adjunct, provided that the PPE and surrogate are adequately represented, but a combined empirical/analytical simulation approach is preferable to overcome the inaccuracies and limitations inherent to both methods.

Secondly, during the new product evaluation phase it is arguably unethical to rely on computational analysis alone before human-use trials are conducted. Superior human surrogates that enable manufacturers to evaluate physical prototype PPE performance, without the risk of human injury, are needed to test the effectiveness of ‘what has been made’ rather than relying on ‘what was intended or predicted.’ The PPE industry, including the sporting goods sector, recognises that physical human surrogates are essential in this second phase.

1.2. Aims and Objectives

Given the shortcomings associated with rigid surrogates used in sports PPE testing standards and the potential inaccuracies introduced through such measures there is a clear need for improved surrogates. A more biofidelic human body surrogate can potentially provide a better representation of the sports impacts experienced and provide a more accurate prediction of the effectiveness of PPE and the likelihood of injuries. Improved surrogate technologies have potential further applications in military ballistic impacts, automotive vehicle and pedestrian impacts, and medical training dummies.

The materials used in a synthetic human surrogate significantly affect their behaviour. It is important to match the dynamic stiffness of the human tissue structures to provide a biofidelic impact response (Saraf *et al.*, 2007). Silicone rubbers have been previously used as a soft tissue simulant (e.g. Hrysomallis, 2009) due to their durability, consistency of response and comparable material density (approximately 1000kg.m^{-3}). A previous study, conducted by Payne *et al.* (2014) illustrated the benefits of a bespoke relaxed muscle tissue simulant when compared to Dow Corning Silastic 3481 single material simulant. The study showed the simulant produced a more biofidelic response under sports specific impact conditions in a simple puck geometry when using quasi-static material data to populate the material models.

Although previous sports impact surrogates have employed a single soft tissue material to represent all of the human structures (excepting bone) in a body segment, treating them as a homogenous block, it is believed that this approach fails to reproduce important interactions between different tissue structures such as variable stiffness, relative movement between structures, pressure distribution and deformation of tissues distant from the impact site. It is hypothesised that a structured multi-material soft tissue surrogate will provide a more biofidelic representation of human tissue response than single material constructions. The human body segments pertinent for this multi-layered soft tissue design are large fleshy areas surrounding a bone such as the thigh, upper arm and lower leg regions.

The dominant soft tissue in these fleshy regions is muscle. A further criticism of previous soft tissue surrogate materials is that they fail to reproduce the anisotropic behaviour of real muscle. There is a growing body of literature concerned with quantifying this aspect of muscle tissue behaviour (§2.1) but achieving consensus and a complete description of human muscle behaviour *in vivo* loaded beyond quasi-static strain rates, that would enable the development of suitable anisotropic surrogate materials, is arguably some years away. In the meantime the sports and other PPE industries could benefit from improved physical surrogates based on better isotropic material combinations. Success in producing these intermediate surrogates arguably paves the way to the surrogates of the future containing synthetic anisotropic muscle tissue, particularly if these anisotropic materials are based on modified isotropic counterparts.

In this study, the development of improved relaxed and contracted muscle tissue simulants are presented for use in a superior multi-tissue human surrogate limb segment subject to blunt trauma from compressive impact. Although the ultimate goal of such simulants is to match all

of the non-linear, anisotropic, viscoelastic material behaviour exhibited by organic muscle tissue a multi-layered surrogate based on isotropic materials represents a significant advancement in biofidelity from previous surrogate designs.

To this end a detailed description of the fabrication of additive cure silicones has been documented for both relaxed and contracted muscle tissue simulants. The fabrication of such simulants has been explored and the mechanical properties of the materials in uniaxial compression investigated through a range of representative strain rates. A range of commonly used constitutive models (Neo Hooke, Mooney-Rivlin and Ogden) have been fit to experimental data to describe their hyperelastic material behaviour. A Prony series expansion was also generated to describe their viscoelastic behaviour.

2. Surrogate Materials Development

2.1. Structure and Composition of Organic Muscle Tissue

Skeletal muscle constitutes the bulk of soft tissue in the human limb segments, accounting for 40-50% of body mass (Wang *et al.*, 1997; Salem *et al.*, 2006), and as such is important to represent effectively. The microstructure of skeletal muscle tissues are characterised by a hierarchy of densely packed muscle fibres arranged in fascicles surrounded by connective tissues. Muscle tissue is heterogeneous and exhibits anisotropy, viscoelasticity, and strain rate dependencies.

It is very challenging to practically obtain *in vivo* muscle human data from mechanical tests and as such most studies are conducted *ex vivo*. There is also a paucity of good quality research in human tissue properties due to difficulties in obtaining post-mortem human subjects and the degradation in properties associated with the conditions the body is kept post-mortem prior to testing (Van Ee *et al.*, 2000) and storage conditions (Menz, 1971; Lee & Pelker, 1985; Clavert *et al.*, 2001). Consequently, very few studies presented are of human tissues (Chawla *et al.*, 2009; Balaraman *et al.*, 2012). Much more of the muscle characterisation research published being obtained from testing on more accessible organic animal specimens such as bovine (McElhaney, 1966; Van Sligtenhorst *et al.*, 2006) or porcine sources (Song *et al.*, 2007; Van Looke *et al.*, 2006, 2008, 2009). It is suggested that these tissues exhibit anatomically and biologically similar structures to humans (Douglas, 1972; Snedeker *et al.*, 2005). Although, Balaraman *et al.* (2012) reported that human muscle tissues studied were stiffer than both bovine and porcine equivalents.

Many other factors have been reported to affect the mechanical properties of characterised tissues:

- Age of the specimen: Song *et al.* (2007) noted a difference in mechanical response compared with the study conducted by Van Looke *et al.* (2006) and partly attributed differences to the age of the specimens (3 and 5 month old animals respectively).
- Location of the specimen: Balaraman *et al.* (2012) observed significantly greater stiffness in tests conducted on *in vitro* human gastrocnemius muscles compared to soleus muscles when tested at comparable strain rates.

- Directionality of testing: Several studies have acknowledged that porcine tissues behave stiffer when tested perpendicular to the direction of the muscle fibres (Van Loocke *et al.*, 2006; Böl *et al.*, 2012; Song *et al.*, 2007).
- Time post-mortem prior to testing: Van Sligtenhorst *et al.* (2006) showed that there was a significant difference in the mechanical properties of specimens 1.5 and 3.5 hours post-mortem when compared to specimens 5,6,7,9 and 27 hours post-mortem; this was largely supported by Van Loocke *et al.* (2006) who suggested that dehydration of tissues may lead to an increased stiffness.
- Shape of specimen: Inaccuracies may be introduced due to the specimen shape; it was reported that it was almost impossible to cut perfect shapes due to the high mobility of fresh muscle tissues (Böl *et al.*, 2012; Van Loocke *et al.*, 2006).

Many characterisation studies have been conducted investigating the mechanical properties of muscle tissues. Most tests are conducted in uniaxial compression and have been performed under quasi-static loading conditions (Yamada, 1970; Fung 1993), which are unrepresentative of the dynamic loading experienced from sports impacts. McElhaney (1966) conducted a seminal study characterising *in vitro* bovine tissues around the femur in uniaxial compression using both a Tinius-Olsen electromatic test machine and an air operated test machine for tests at low and high strain rates respectively. In recent years, authors have typically investigated the high strain rate response of soft tissues using a polymeric Split Hopkinson Pressure Bar (SHPB), capable of testing in the order of 10^3 s^{-1} (Van Sligtenhorst *et al.*, 2006; Song *et al.*, 2007; Chawla *et al.*, 2009; Balaraman *et al.*, 2012). There is, however, a paucity of information conducted under intermediate strain rate conditions in a similar range to those expected in sports impact scenarios (approximately $50\text{-}500 \text{ s}^{-1}$), based on a maximal velocity projectile impact from a cricket ball of 34.6 ms^{-1} (Penrose *et al.*, 1976).

Engineering stress-strain data from a representative subset of studies are presented in Figure 2.

Figure 2 – Compressive engineering stress-strain plot showing data from previous organic tissue characterisation studies

The muscle property data, however, lacks tonicity that would be present in live humans; a very different response may occur when an athlete's limb segment is contracting, braced or tensed reflexively in anticipation of an impact. Although, it is widely suggested that muscles exhibit a greater stiffness when contracted (Hoffer & Andreassen, 1981), there is no consensus quantifying the effect that muscle tonicity introduces. Many different studies have investigated the effect of contraction using a variety of techniques such as ultrasound and motion capture (Blackburn *et al.*, 2009), magnetic resonance elastography (Bensamoun *et al.*, 2006), vibro-ultrasound (Wang *et al.*, 2012) and manual indentation and ultrasound (Zheng *et al.*, 1999); the effects measured varied. Crisco *et al.*, (1996) also observed a 44% reduction in the delay between peak force and peak displacement between contracted and relaxed muscles when testing Wistar rats *in vivo*; this is an indication of reduced viscoelasticity in contracted muscles. Given no definite characterisation of this effect, a direct measurement by Zheng *et*

al. (1999) was used to provide an approximation of the potential stiffness of a contracted skeletal muscle. The study used a manual ultrasound indentation system on the human forearm and lower limb soft tissues and recorded a mean increase in effective Young's modulus of 4.2 times when gripping a pen. Without further data to better inform this study, a uniform scaling factor of $4.2\times$ the relaxed muscle stress-strain data was employed.

Given the large degree of variation between characterisation studies based on the specimen type and nature of mechanical loading (Fig. 2), it is clear that there is no consensus on the definitive structural response. Consequently, it is to be noted that this study aims to provide a close approximation of a representative target dataset, demonstrating the possibility of matching the properties of any particular set of organic tissue data (partly in anticipation of better definition of human tissue response in the future). The mechanical properties of the organic muscle tissue from uniaxial compression tests in the cross-fibre direction have been selected as target properties to mimic with silicone simulants. The materials have been characterised in compression to match the mechanical properties of muscle in the cross-fibre direction most pertinent to blunt force trauma in the aforementioned body regions of interest. Approximating the behaviour of an anisotropic organic material with a tailored isotropic one should yield a useful result when the anticipated loading of the simulant corresponds with the characterisation conditions for the organic tissue.

The organic tissue responses reported by McElhaney (1966) and Song *et al.* (2007) taken in the cross-fibre direction are nearly coincident in their overlapping strain ranges at both quasi-static and intermediate strain rate ($100\text{-}540\text{s}^{-1}$) ranges. Consequently, the stress-strain data were combined to a wider strain range (appropriate to blunt trauma from compressive impacts in sport). The scaled responses of contracted muscle tissue are also presented in Figure 3.

Figure 3 – Engineering stress-strain graphs showing organic tissue properties from relaxed (McElhaney, 1966; Song *et al.*, 2007) and contracted (Zheng *et al.*, 1999) muscle tissue in (a) linear and (b) log-linear plots

The success of the developed soft tissue simulants have been considered in relation to the following questions:

- Given current *in vivo* and *ex vivo* data, do the developed PDMS simulants exhibit a more biofidelic response than previously used single material simulants?
- In the future, given superior *in vivo* material properties, is there scope to use the same approach to produce an improved PDMS formulation?
- Do the PDMS simulants perfectly match the organic tissue characteristics?

2.2. Formulations

Two-part (A and B) additive cure vinyl blocked Polydimethylsiloxane (PDMS) silicones (Technovent Ltd., Bridgend, UK) were formulated through an iterative development process altering polymer and cross-linker constituent concentrations. It is possible to use, in different proportions, polymers of different chain lengths and so molecular weights. These cross-link within the silicone rubber by reacting at the end vinyl groups to form a cross-link network.

Optimal cross-linking of longer polymer chains produces a more flexible elastomer; conversely, the use of a greater proportion of cross-linked shorter polymer chains produces a more inelastic/brittle polymer. In addition silica fillers are used to stabilise and reinforce the cross-linked matrix, adding strength but also stiffness to the resulting PDMS silicone.

In this study the part A component consisted of three different chain length polymers with vinyl termination ($\text{CH}=\text{CH}_2$): V21 (molecular weight: 6000 Daltons), V31 (molecular weight: 28,000 Daltons) and V46 (molecular weight: 117,000 Daltons). The concentrations of these polymers were varied to gain desired material properties.

Each polymer had 20% (by mass) surface treated (hexamethyldisilazane treated) silica filler concentration to stabilise and reinforce the cross-linked matrix and increase the strength of the polymers. A 10% (by mass) concentration heat-cure platinum catalyst (platinum cyclovinyldimethylsiloxane complex) was also present in the Part A solution, which permitted significant working time, requiring the application of heat to burn off the masking compound.

The Part B component consisted of a MethylHydroxiloxane-Dimethylsiloxane Copolymer (H301), which was used to chemically bond the polymer chains via a hydrosilation addition reaction. This was mixed in a 1:9 solution (by mass) with an unfilled V21 polymer acting as carrying agent for the cross-linker. The Part A:B ratios in the solutions were varied to alter the cross-linker concentration and thus change the material properties. A greater concentration of cross linker increases the stiffness of resultant silicones (Gelest, 2014).

Deleted: ; a greater concentration of cross linker was shown to increase the stiffness of the silicones.

Figure 4 – Quasi-static compressive engineering stress-strain plot of silicones with different cross-linker concentrations in: (a) linear and (b) log-linear graphs

The chemical structure of the additive cure reaction is shown in Figure 5.

Figure 5 - Chemical structure of additive cure PDMS silicone reaction

2.3. Fabrication Procedures

The process by which the silicones were prepared may significantly affect the mechanical properties of the cured elastomer. Factors such as weighing inaccuracies, method of mixing, time left before curing and the temperature of cure can all alter the quality of the resulting silicones. The process employed in the silicone development is outlined in below:

1. Weighing of constituents ($\pm 0.1\text{g}$ accuracy)
2. Constituents poured and mixed manually for 2-3 minutes
3. Silicones vacuumed for 20-30 minutes until fully degassed.
4. Moulds coated in several layers of release wax and Polyvinyl Acetate (PVA) release agent.
5. Solutions poured into specimen moulds immediately after vacuuming.
6. Specimens heated at 60°C in an environmental chamber until cured (approx. 30 minutes).

7. Cured specimens removed from moulds and cleaned from contaminate left by the release agents using a standard commercial cleaning solution.

The bespoke silicone blends developed for the muscle tissue simulants were formulated through an iterative process:

- A range of different 'Part A' formulations with V46:V31:V21 ratios from 85%:10%:5% to 52%:24%:24% were combined in a range of Part A:B ratios from 10:1 to 1:1
- Uniaxial compression testing under quasi-static loading rates was used to evaluate material samples against the target organic tissue dataset.
- Further incremental formulation adjustment was conducted on blends indicating a promising match.

The final formulations are presented in Table 1.

Table 1 - Part A and B silicone constituent concentrations (% by mass)

3. Mechanical Testing Procedures

3.1. Background

To gain a comprehensive understanding of material behaviour both stress-strain data and viscoelastic material properties are required. It is also important to characterise the material at strain rates representative of those experienced by the surrogates in sports impact scenarios. Muscle tissues have been demonstrated to exhibit both large strain deformations *in vivo* and high strain rate dependencies when subject to impact stimuli (Van Looke *et al.*, 2006; Song *et al.*, 2007). Given the nature of many sports impacts, high strain rate, high strain deformations are also likely and as such require material characterisation at comparable rates to ensure consistency of response and adequately populate associated parallel Finite Element (FE) models.

Silicone elastomers typically exhibit a high strain rate sensitivity, which is attributed to the timescale required for polymer chains to re-orientate during deformation. During deformation at low strain rates the polymers have significant time to reorient, whilst at high strain rates the deformation of polymer chains is restricted to bending and stretching of chemical bonds (Shergold *et al.*, 2006). Many high strain rate characterisation procedures use an oscillatory dynamic test machine (e.g. Instron Electropuls), which can test materials at high strain rates but compromises the degree of mechanical strain (e.g. Arbogast *et al.*, 1997; Van Looke, 2009). These test machines also introduce ringing phenomena, inertial effects and inaccuracies from the elastic response of the machines influencing the stress-strain outputs.

Filled silicone elastomers are also widely reported to exhibit the Mullins effect (Mullins, 1969; Yan *et al.*, 2012; Rey *et al.*, 2013) and this has therefore been studied to examine the extent of their effect in the developed materials. The presence of such an effect has been used to understand the steady state material behaviour, where the loading curves become almost coincident following repeat loading cycles, and was subsequently used to populate material

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models. This information could also potentially inform the manner in which physical synthetic surrogates are handled prior to testing and whether preconditioning may be necessary to ensure consistency of response.

3.2. General Procedure

A series of unconfined uniaxial compressive tests were conducted using ASTM D395 cylindrical material specimens (29mmØ, 12.5mm height). For each of the specimens, the ends were coated with a Vaseline lubricant to reduce plate-specimen friction and barrelling effects, which can account for up to a 50% error in compressive stress (Wu *et al.*, 2004). All tests were conducted at room temperature (22°C), within the guidelines of $23 \pm 2^\circ\text{C}$ stipulated by ASTM D1349 for rubber testing. Humidity was not controlled in the experiments.

3.3. Low Strain Rate Tests

Quasi-static strain rate tests (0.4s^{-1}) were conducted with an Instron 5569 screw driven universal test machine using a range of load cells between 100N and 50kN sensitivity dependent on the specimen tested and a linear variable differential transformer (LVDT) to determine displacement with a sampling rate of 100Hz. Each specimen was placed directly between flat compressive platens and aligned to ensure planarity and parallelism; the load cell and LVDT were zeroed prior to testing. The software was programmed to perform a triangular cyclic compressive protocol to ensure a constant strain rate was maintained in loading and unloading cycles. The specimen was compressed incrementally through six loading/unloading cycles in 0.1 compressive strain intervals until failure (Fig. 6). This incremental procedure was adopted to investigate the presence of the Mullins effect in the material following previous guidelines and usage by Dorfmann & Ogden (2004) and Diani *et al.* (2009).

Figure 6 - Programmed triangular loading profile for quasi-static uniaxial compressive tests

The Poisson's ratio was an important parameter to measure as only compressive loading tests were conducted; this ratio is the sole determinant of volumetric change in the material and therefore the behaviour in other mechanical loading directions. The deformations were measured using a high speed imaging camera aligned perpendicular to the specimen and captured at 2000 frames per second (fps).

3.4. Intermediate Strain Rate Tests

An Instron 9250 drop tower was used to test the material to both intermediate strain rates and high strains. The test machine is equipped with three piezoelectric load cells beneath the bottom compressive platen. A high speed, high resolution video camera was used to measure specimen displacement. The camera had a 256k greyscale image resolution of 1024x1024 pixels ($\approx 100\text{mm} \times 100\text{mm}$), shutter speed of 1/10000s and was operated at a frame rate of 4000fps to provide sufficient spatial and temporal resolution to accurately measure changes in strain. A 6.8kg mass with a flat compressive platen end effector was dropped from a series

of heights (0.01m-0.5m) to induce high strain at a range of strain rates in the material. The strain rates reported are characterised by the initial maximum value recorded.

3.5. Stress Relaxation Tests

Uniaxial ramp-and-hold stress relaxation tests were performed on the Instron 5569 to understand the viscoelastic time-dependant properties of the material. The compressive strain levels induced in test specimens vary across standardised tests (ISO 3384-1:2011; ASTM D6147-97:2008). The strain chosen was the highest strain attained by all materials before failure (established previously) was selected to characterise the materials under the most representative conditions for compressive sports impacts. Each specimen was deformed to 0.5 strain at a rate of 0.4s^{-1} and held under constant strain for a period of 300s. The 300s time period was used previously by Van Looke *et al.* (2008) and was shown to be sufficient for the materials to achieve a steady relaxed state.

3.6. Analysis Procedures

Force-time and displacement-time data were obtained from the mechanical tests. The engineering strain (ϵ) at each time interval was calculated with Equation (1) using displacement values and the original length measured prior to testing using a set of Vernier callipers ($\pm 1 \times 10^{-5}\text{m}$). Engineering stress (σ) was calculated (Equation 2) using data from the load cell divided by an original cross-sectional area of $6.6 \times 10^{-4}\text{m}^2$.

$$\epsilon = \frac{\Delta l}{l_0} \quad (1)$$

Where: l_0 = original length; Δl = change in length.

$$\sigma = \frac{F}{A_0} \quad (2)$$

Where F = applied load; A_0 = original cross-sectional area

There is currently no consensus as to the best method to quantify the extent of the Mullins effect (Diani *et al.*, 2009). In this study, the maximal stress reductions from the loading curve of the first to last cycle at maximal strain were calculated.

The Poisson's ratio (ν) was determined through post processing of the imaging output using video analysis software Image Pro Analyser (Media Cybernetics Inc.) to determine the relative ratio of transverse and axial strains (Eq. 3).

$$\nu = -\left(\frac{\Delta \epsilon_{trans}}{\Delta \epsilon_{axial}}\right) \quad (3)$$

Where: $d\epsilon_{trans}$ = change in transverse strain; $d\epsilon_{axial}$ = change in axial strain

The intermediate strain rate drop tests were post-processed using the load cell data and positional output from video analysis. The loading curves from each dataset were extracted

and used to calculate compressive stress-strain parameters at each time increment. The data was then smoothed using polynomial fits to remove noise introduced from the load cell output (Fig. 7).

Figure 7 - Processing of intermediate strain rate drop tower data: (a) raw data; (b) smoothed stress-strain plot

4. Results

4.1. Low Strain Rate Tests

Engineering stress-strain results for both relaxed and contracted muscle simulant materials (Table 1) are shown in Figure 8a and 8b in comparison with the published organic tissue data (cf. Fig. 3). The graphs illustrate that both simulant materials show responses in the same stiffness range as organic tissues, however, the nature of the responses differ. The greatest divergence was present in the contracted muscle simulant at low strains with an error of up to 250%. The silicone simulants typically initially exhibit a stiffer response up to approximately 0.4 strain at which point the organic tissues stiffen by a greater magnitude than the silicones.

The responses of the Silastic 3481 simulant were also compared to the target organic tissue datasets under quasi-static strain rates (Fig. 8c, 8d). The simulant provided a poor approximation of relaxed muscle simulant with errors at low strains of over 4000% but showed a more comparable fit to the contracted muscle tissue simulant. The Silastic simulant followed the same stress response as the PDMS silicones with a weaker strain hardening being exhibited at approximately 0.4 strain.

A direct comparison of the closeness of fit of the contracted PDMS silicone and Silastic 3481 simulants with the target organic tissue property dataset is shown in Figure 8e. The graph shows that both silicone simulants exhibit a far greater stress response compared to organic tissue data at low strains. This increased stiffness is most pronounced in the Silastic 3481 simulant which exhibited a significantly greater compressive stress response below 0.1 strain with up to 896% increased stiffness.

Figure 8 – Quasi-static stress-strain plots showing the divergences in compressive response between: (a) PDMS relaxed muscle simulant and organic tissue data; (b) PDMS contracted muscle simulant with organic tissue data; (c) Silastic 3481 with relaxed organic tissue data; (d) Silastic 3481 with contracted muscle organic tissue data; (e) Percentage errors plots showing divergence between silicone simulants and target organic contracted muscle data

Mullins Effect

The percentage difference in compressive stresses between the first and last loading cycles at maximal strain were recorded and displayed in Figure 9.

Figure 9 - Engineering stress-strain plot showing stress softening effect in cyclic compressive protocol between the first and last cycles in: (a) PDMS relaxed muscle simulant and (b) PDMS contracted muscle simulant

The results show a far greater stress-softening effect was exhibited in the softer relaxed muscle simulant with a reduction in maximal stress of 12% compared to 0.7% recorded for contracted muscle under identical loading conditions. Both tissue simulants exhibit a significant hysteresis loss in the unloading cycles demonstrating their viscous behaviour.

Poisson's Ratio

The measured values of Poisson's ratio for relaxed and contracted muscle simulants were 0.493 and 0.480 respectively. Both simulants are shown to be virtually incompressible, in agreement with the literature concerned with elastomers (Ali *et al.*, 2010; Mottahedi *et al.*, 2011) and organic muscle tissue (Van Looke *et al.*, 2009; Böl *et al.*, 2012). As reported elsewhere (Shergold *et al.*, 2006), compressive testing of incompressible materials conveniently incorporates the biaxial tensile response of the specimen.

4.2. Intermediate Strain Rate Tests

The response of the materials tested at a range of strain rates are shown in Figure 10. The results show an increase in the material stiffness of both simulants at increased strain rates.

Figure 10 – Intermediate strain rate engineering stress-strain curves for: (a) PDMS relaxed muscle simulant and (b) PDMS contracted muscle simulant

4.3. Stress Relaxation Tests

The magnitude of stress relaxation in the materials is shown in Figure 11 when held under a constant strain. The softer relaxed muscle specimen exhibited a far greater stress reduction.

Figure 11 Normalised stress-time plot showing stress relaxation in PDMS relaxed and contracted muscle simulants

4.4. Dynamic Response Comparisons

A previous study conducted by Payne *et al.* (2014) illustrated the benefits of the relaxed muscle tissue simulant when compared to Silastic 3481 when using quasi-static material properties. Further comparison is made in Figure 8 between the compressive stress-strain response of organic tissues (at quasi-static and intermediate strain rates) and Silastic 3481. Figure 8 also shows the stress-strain response of the PDMS relaxed and contracted muscle simulants (at similar strain rates). The banded regions show the response corridors for the PDMS simulants through the given range of strain rates.

Figure 12 – Log-linear engineering stress-strain plots showing a comparison between Silastic 3481, PDMS silicones and organic tissue properties for: (a) PDMS relaxed muscle simulant and (b) PDMS contracted muscle simulant

Figure 12 clearly shows that Silastic 3481 is much stiffer than organic tissues at both quasi-static and intermediate strain rates. The PDMS simulants provide a much closer representation of the organic tissue behaviour at low strain rates (especially the relaxed muscle simulant). At increased strain rates the PDMS simulants exhibit a far better representation of target organic tissue behaviour than Silastic 3481 but are unfortunately less than ideal.

5. Constitutive Modelling

Silicone elastomers exhibit both elastic and viscous resistance to deformation as the materials retain the recoverable strain energy partially but dissipate energy if the deformation is maintained. Consequently, combinations of hyperelastic and viscoelastic models are needed to represent the non-linear and time-dependant behaviours of elastomeric materials.

6.1. Hyperelasticity

The simulant materials all exhibit significant strain hardening effects and as such require hyperelastic models to describe their stress-strain behaviour. First order phenomenological models were evaluated which consider changes in stiffness based on the strain energy density (W) of the material. The Ogden, Mooney-Rivlin and Neo Hookean have been widely used in literature to describe incompressible, hyperelastic materials (Marckmann & Verron, 2006; Ali *et al.*, 2010) and were evaluated for each material. The Ogden model (Ogden, 1972) is a commonly used hyperelastic model to describe the material behaviour of soft solids that are both isotropic and incompressible. The model proposes the strain energy function (W) is based on the principal stretches ($\lambda_1, \lambda_2, \lambda_3$), with μ describing the shear behaviour and α describing the strain hardening (Equation 3). The Mooney-Rivlin model (Rivlin, 1948) is another commonly used material model to describe rubber-like behaviour due to its simplicity and ability to model moderate deformations and high strain hardening (Shergold *et al.*, 2006; Meunier *et al.*, 2008). It is a first order polynomial model represented by Equation (4) where C_{10} is a material parameter and $C_{00}=0$. The Neo-Hookean model (Treloar, 1943) is a first order reduced polynomial. It is the simplest hyperelastic constitutive equation for describing rubber behaviour with only one material parameter C_{10} , and is often used when material data is insufficient (Marckmann & Verron, 2006).

Ogden:
$$W = \frac{2\mu}{\alpha^2} (\bar{\lambda}_1^{\alpha_1} + \bar{\lambda}_2^{\alpha_1} + \bar{\lambda}_3^{\alpha_1} - 3) \quad (4)$$

Where: λ_i = principal stretch ratios; μ, α = material constants

Mooney-Rivlin:
$$W = \sum_{i,j=0}^N C_{ij} (\bar{I}_1 - 3)^i (\bar{I}_2 - 3)^j \quad (5)$$

Where: $\bar{I}_1, \bar{I}_2$ = stress invariants; C_{ij} = material constants;

Neo-Hooke:
$$W = C_{10}(\bar{I}_1 - 3) + (\bar{I}_2 - 3) \quad (6)$$

Typically, FE solvers determine material parameters automatically by fitting a strain energy function to input uniaxial test data based on a non-linear least squares optimisation. A commercial FE solver, Abaqus Explicit (Version 6.13) was used to perform this analysis; the quality of the fits are shown in Figure 13 with generated material coefficients presented in Tables 2-4 alongside RMS errors showing the degree of divergence of each dataset from the experimental data.

Figure 13 – Hyperelastic model fits for: (a) PDMS relaxed muscle simulant and (b) PDMS contracted muscle simulant at a range of intermediate strain rates

The Mooney-Rivlin model typically provided a good fit ($< 0.229\text{MPa}$ RMS error) for both simulants through the range of strain rates reported. The model, however typically underestimated the magnitude of strain hardening at high strains (e.g. relaxed muscle, 112s^{-1}). The Ogden model also generally provided a good fit ($< 0.730\text{MPa}$ RMS error) through the

range of strain rates tested but typically under predicted the compressive stresses under low strain, low strain rate conditions and over predicted the strain hardening present at higher strains (>0.5 strain). The Neo Hooke model provided the worst performance of the three models and was unable to capture much of the non-linearity of the silicone behaviour, particularly in the stiffer contracted muscle simulant (e.g. contracted muscle, 50s^{-1} and 194s^{-1}).

Table 2 - Mooney-Rivlin model coefficients for PDMS simulants

Table 3 - Ogden model coefficients for PDMS Simulants

Table 4 - Neo Hooke model coefficients for PDMS simulants

5.1. Viscoelastic Models

A generalised Maxwell model known as the Prony series (Tschoegl, 1989) was used to model the viscoelastic properties of the simulants. The Prony series models a series of Maxwell models in parallel and considers the relaxation of the materials through a series of time intervals. The force-time data from the stress relaxation tests were manipulated to determine the compressive stresses at each time interval. The uniaxial stresses were then normalised from the point of maximal stress to determine the shear modulus parameter, $g(t)$ (Equation 7). The materials are assumed to be incompressible and as such the bulk modulus parameter, $k(t)$ is considered to equal zero.

$$g(t) = \frac{\sigma(t)}{\sigma_0}, \quad \sigma_0 = \sigma(0) \quad (7)$$

$$g(t) = 1 - \sum_{i=1}^N g_i (1 - e^{-t/\tau_i}) \quad (8)$$

The normalised shear modulus values were then used to generate time-dependent Prony series parameters using equation (9) in the automated feature of the FE solver fitting a curve to the relaxation data. The quality of the curve fits are shown in Figure 14 and the Prony series coefficients describing the viscoelastic behaviour of both simulants are shown in Table 5.

Figure 14 - Prony series model fits for: (a) PDMS relaxed muscle simulant and (b) PDMS contracted muscle simulants

Table 5 - Prony series coefficients for PDMS simulants

6. Discussion

6.1. Organic Muscle Tissue Data

Previous studies investigating the mechanical behaviour of muscle tissues have reported a wide range of responses when tested in compression. This is due to many factors relating to

the specimen type and loading conditions utilised. Consequently, to provide an estimation of the general response of such tissues, representative data from studies by McElhaney (1966) and Song *et al.* (2007) were extracted and combined (Fig. 2). Response data characterised from studies conducted in the cross-fibre direction was utilised as an initial simplification of the anisotropic behaviour exhibited by muscle tissues (Van Loocke *et al.*, 2006; Song *et al.*, 2007; Böl *et al.*, 2012).

A fundamental issue with the organic tissue data, however, is that it has been gathered post-mortem and lacks tonicity. This is always present to some degree in humans due to neurological effects. Much current research has also been conducted at higher strain rates relevant to automotive or ballistic impacts, there currently a paucity of data at intermediate strain rates pertinent to sports impacts. This causes issues in extrapolating outputs to apply to these data ranges. Given no standardised measure for characterising organic tissues (in particular at high strain rates) much of the differences reported could be attributed to variability in the test methods. Studies typically use a SHPB to perform high rate materials characterisation. When experiments are conducted with high stiffness pressure bars errors can be introduced through inertial effects in low-impedance materials, which can often be at the same order of magnitude as mechanical properties making it difficult to determine intrinsic material behaviour (Song *et al.*, 2007). More recent characterisation studies, however, have managed to overcome such issues. For example, a study by Trexler *et al.* (2011) developed practices for use of a modified SHPB with low impedance soft biomaterials using pulse shaping techniques and thin specimen sizes.

The target muscle tissue dataset was obtained from previous studies through a range of unmatched strain rates and loading conditions. A greater confidence in the mechanical behaviour could be obtained through internal testing of organic tissues at strains and strain rates pertinent to sports impacts. This could potentially provide an enhanced target dataset from which to base new formulation of PDMS silicones.

The mechanical properties of contracted muscle tissue remain unknown and almost certainly vary with the levels of muscle contraction. Some previous research has concluded that contracted muscle is less susceptible to injuries than relaxed muscle (Crisco *et al.*, 1996; Tsui & Pain, 2008) but this is not necessarily so. Given further research to better quantify the behaviour of contracted muscle new simulants could be formulated with the methods presented.

6.2. Silicone Behaviour

The developed PDMS silicone simulants each exhibited mechanical hysteresis, stress softening effect and strain rate dependency. Through the range of strain rates tested, the silicones exhibit a significant strain rate dependency whereby at higher strain rates the silicones experience a more pronounced strain hardening effect; this was more apparent in the stiffer contracted muscle simulant. This behaviour is consistent with results from literature (Gray III *et al.*, 1997; Lee *et al.*, 2003; Kakavas, 1996; Jones, 1960). Gray III *et al.* (1997) conducted uniaxial compressive tests on silicone rubbers and noted that the Young's modulus

increased by a factor of 8 when strain rate was increased from 10^{-3} to $3 \times 10^3 \text{ s}^{-1}$, whilst little change was observed between strain rates 10^{-3} - 10^{-1} s^{-1} .

The cyclic compressive tests also showed that the simulants both experienced significant mechanical hysteresis losses, which were more prominent at higher strains. This is indicative of increased viscous behaviour at increased strain rates. The Mullins effect is also a pertinent phenomenon that changes the way in which elastomers behave after multiple deformations. The PDMS silicones each exhibited a Mullins effect, though this was far more pronounced in the softer muscle relaxed simulant. The coincident behaviour of the silicones after multiple deformations indicates that it may be necessary to precondition the specimens to ensure consistency of response. The presence of this effect was consistent with other studies, which have observed a Mullins effect in filled silicones (Yan *et al.*, 2012; Rey *et al.*, 2013).

The mechanical response phenomena experienced are in contrast to results from a study by Meunier *et al.* (2008), where little hysteresis, stress-softening and strain rate dependency were observed. This could be attributed to the lack of filler content in the silicones, which has been suggested to be a strain amplifier (Rault *et al.*, 2006; Podnos *et al.*, 2006; Mullins & Tobin, 1965). The presence of filler particles in the silicone can significantly restrict the mobility of polymer chains and increase the degree of non-linearity exhibited (Cosgrove *et al.* 1997; Cassagnau & Méliis, 2003). It is also suggested that less filled rubber exhibits purely entropic behaviour with no mechanical dissipation, whilst filled rubber exhibits both viscous and entropic behaviour, increasing temperature sensitivity (Martinez, 2013). The effect of temperature on the behaviour of the silicones, although controlled was not examined explicitly in this study and has potential effects on the mechanical response. Rey *et al.* (2013) noted that filled silicones exhibited an increase in stiffness at increasing temperatures which shows that the effects of entropic changes are much greater than viscous changes. A greater stress softening effect was also experienced in filled rubbers at lower temperatures below 60°C . Further manipulation of the silicone formulations may consider the concentration and type of filler present in the polymer to alter the mechanical properties of the cured silicone and create improved parallels with the target organic data.

The variability between equivalent silicone blends was not presented in this study but the quality of blend is believed to affect the resultant stiffness of the material. The PDMS silicones are largely dependent on the consistency of mix, temperature and time of cure all relating to the number of cross-links produced. The values presented are intended to be an indication rather than absolute values for that particular formulation. Further research is required to establish batch sensitivity to minor variations in production variables for different quantities of simulant.

6.3. Constitutive Modelling

Three simple hyperelastic models were evaluated as representative subsets of commonly used material models. The Ogden and Mooney-Rivlin models provided the closest material fits through both simulants and the range of strain rates conducted. The quality of the constitutive model fits are generally in contrast with the findings from Meunier *et al.* (2008) and Shergold

et al. (2006), who observed a weak Mooney-Rivlin fit, though the decreased strain hardening effect was noted when compared to the Ogden model. However, the variability present between hyperelastic fits and the difference of error in each does reiterate the need for caution when interpreting the results from FE models.

The constitutive models could potentially be improved for the simulant materials. Complex hyperelastic models such as the Bergström-Boyce model (Bergström & Boyce, 1998), which model both hysteresis behaviour and strain rate dependency, could provide an improved singular description of the materials. Similarly, incorporating the viscoelastic material properties under higher strain rates could be beneficial. Van Loocke *et al.* (2008) showed the viscoelastic component of muscle tissue increased at higher compression rates. A better understanding of the simulant behaviour under these conditions would be useful and could potentially provide an improved description of the viscoelastic properties.

6.4. PDMS Comparison with Organic Tissue

6.4.1. Are the PDMS formulations better than surrogate materials?

The success criteria of the developed silicone simulants were detailed previously (§. 1.3) and have been outlined again below.

- Is there scope with current *in vitro* data to develop materials that exhibit a more biofidelic response than previously used single material simulants?
- If better *in vivo* material properties become known, is there scope to develop silicones to match the new data?
- Is it possible to develop a silicone simulant that perfectly matches the organic tissue properties?

Using existing *in vitro* material data from literature sources the quality of the PDMS simulant fits were evaluated. Under quasi-static loading conditions, the relaxed muscle tissue simulant exhibited a much improved representation of the organic tissue dataset than Silastic 3481 (Fig. 8a, 8c). A maximal error of 149% was recorded at low strains (< 0.2) beyond which the PDMS simulant was within a 50% error. Silastic 3481 showed a 4086% error in the low strain range and a response typically between 250-500% beyond this point.

The contracted muscle tissue simulant showed a greater divergence from the organic tissue dataset and generally presented a broadly comparable response to the Silastic 3481 simulant (Fig. 8b, 8d). A significant improvement, however, is apparent at low strains (< 0.1) where the Silastic simulant showed an 896% error compared to a 250% error in the PDMS simulant at the same strain (Fig. 8e).

At increased strain rates, the PDMS silicones continued to exhibit an improved representation of the organic tissue behaviour than the Silastic simulant (Fig. 12). The silicones, however, typically exhibited a greater stiffness than the organic tissues at higher strain rates; this response was most pronounced in the relaxed muscle simulant. A large divergence was also present in both simulants between 0.1 and 0.3 strain, where the organic tissue exhibits softer behaviour.

Overall, the silicones exhibited an increase in maximal stress response of $12.9\times (0.4-112\text{s}^{-1})$ and $8.3\times (0.4-194\text{s}^{-1})$ at 0.5 strain in relaxed and contracted muscle tissues respectively. Previous skeletal muscle tissue characterisation studies have generated varied findings. Van Loocke *et al.* (2008) observed a $5\times$ increase in maximal stress between 0.05s^{-1} and 10s^{-1} , whilst Van Loocke *et al.* (2009) observed an increase of $4.1\times$ from $5-3200\text{s}^{-1}$ at 0.3 strain. Similarly, Chawla *et al.* (2009) observed a $1.7\times$ increase in maximal stress between 554s^{-1} and 1001s^{-1} at 0.2 strain. In contrast, Van Sligtenhorst *et al.* (2006) observed an increase of maximal stress of approximately $17\times$ between 0.01s^{-1} and 2300s^{-1} at 0.8 strain and Song *et al.* (2007) observed an $11\times$ increase in maximal stress response between 0.007 and 3700s^{-1} at 0.5 strain. The differences in magnitude appear to be largely indicative of the level of compressive strain applied and as such would suggest that the silicones are behaving in the correct response corridor at higher strain rates up to 0.5 strain.

In general, the silicone simulants exhibited a significantly stiffer response at low strains compared with the target organic tissues. This was a trend across all silicones (Silastic 3481 and PDMS) and is likely to indicate an innate difference in the mechanical behaviour with organic tissues, rather than issues with the quality of the formulations. However, although the primary target dataset is based on porcine and bovine sources, at low strain and high strain rates the formulated PDMS relaxed muscle simulant is conveniently a close match to Chawla (2009) and Balaraman *et al.* (2012) human tissue response data (Fig. 15).

Figure 15 – Intermediate strain rate PDMS relaxed muscle simulant responses compared with human muscle characterisation studies by Chawla *et al.* (2009) and Balaraman *et al.* (2012)

The viscoelastic properties of each simulants have also been compared to previous organic tissue studies using *in vitro* relaxed muscle tissue in the cross-fibre direction (Van Loocke *et al.*, 2008). The normalised shear compliances for each simulant are compared in Figure 16.

Figure 16 - Normalised stress-time plots showing differences in stress relaxation between organic muscle tissue (Van Loocke *et al.*, 2008) and the PDMS relaxed and contracted muscle simulants

Figure 16 shows that the organic muscle tissue exhibits significantly different stress relaxations when compared to the silicone elastomers with the magnitude of reduction after relaxation differing by 48.8% and 54.7% in relaxed and contracted simulants respectively. This could potentially be attributed to the *in vitro* nature of the organic stress relaxation experiment and the re-distribution or expulsion of fluid during a ramp and hold test. The viscoelasticity deficit in the PDMS simulants could be theoretically improved in further iterations of the formulations through improving the viscous behaviour of the isotropic PDMS. However, it may be beneficial to consider material formulations that simultaneously introduce anisotropy and viscous behaviour.

6.4.2. Can the same approach be used to match a better organic tissue data set?

Given the range of stiffness made possible by altering cross-linker concentrations (Fig.4). The range of viable polymer and cross linker ratios outlined in §2.2 can be used to further refine properties, and given the levels of agreement achieved matching two different sets of *in vitro* data in this study, it is evident there is potential for the PDMS silicone formulations to be manipulated to match new *in vivo* data to a similar level of success especially if not too dissimilar.

6.4.3. Can the current PDMS simulants perfectly match organic tissue?

It is not believed that it is possible with PDMS silicones in their current form to perfectly match a given non-linear set of organic tissue stress-strain data. As previously discussed silicones typically exhibited a greater stiffness at low strains followed by a reduced magnitude of strain hardening at higher strains. The silicone elastomers have also been shown to exhibit an increased stiffness at higher strain rates than organic tissues. These innate differences indicate that the PDMS silicones in their current form could not provide an exact match of a particular set of organic tissue properties.

6.5. Applications and Further Study

The presented PDMS silicones have demonstrated capabilities to match target organic tissue datasets and showed improvements in response when compared with previously used surrogate materials. The primary application of these materials is in human impact surrogates, where their merit is in providing an improved representation of overall human impact response. The simulants are intended for use in surrogates representing fleshy areas of the human body such as the thigh, lower leg or upper arm. These structures are also primarily developed to represent blunt trauma normal to the direction of impact and are designed to be good at approximating the net effects of soft tissues in the regions. Conversely, the surrogate structures would not be suitable for modelling injury scenarios such as penetrating impacts, oblique impacts and adverse joint kinematics.

Although the simulants have been characterised extensively, their ultimate response and relative merits remain undetermined until tested, computationally or experimentally, in a surrogate model. A key area for further development of these technologies is through fabrication of bulk synthetic models which would improve understanding of their mechanical response when subject to a wider range of mechanical loading introduced by sports specific impact conditions. Furthermore, the simulants have only been tested and compared to target organic tissue datasets in compression. The anisotropic properties of silicones and behaviour under different loading modes (e.g. predominantly tensile, shear) have not been investigated in this study and represent an area for further study. The development of such simulants could potentially enhance experimentation of different injury types not suited to isotropic materials (e.g. muscle strains, tears).

In addition, the development of skin and subcutaneous adipose tissue simulants are believed to be important in the construction of a more anatomically accurate multi-material soft tissue

surrogate. Although, it is unclear as to whether PDMS silicones will be capable of matching the mechanical properties of very soft (adipose) or very stiff (skin) tissues, current indications from their use show that they are versatile and can be easily manipulated through changes in polymer, cross linker and filler concentrations. The composite response of these tissue simulants collectively will be important to provide an improved global biofidelity and measure of assessment of injury risk and PPE effectiveness.

7. Conclusions

As hypothesised in the introduction (§1.3) a multi-material human soft tissue surrogate for fleshy body segments represents a significant improvement on existing single material surrogates. An initial simplification of such an approach and an intermediary stage of complexity involve the development of soft tissue simulants considering the mechanical behaviour of organic tissues in a single direction. The presented study illustrates the potential of using bespoke blends of additive cure PDMS silicones to match the mechanical properties of relaxed and contracted muscle tissues in uniaxial compression. A subset of representative organic muscle tissue properties were selected as a target dataset from which to compare the PDMS simulants against. Uniaxial compressive tests indicated that under quasi-static loading conditions the relaxed muscle simulant provided a significantly improved response when compared with Silastic 3481. The new contracted muscle simulant showed a large improvement over Silastic 3481 at low strains, but comparable responses at increased strains.

At higher strain rates the responses of the tissues were a poorer representation of the target dataset, where the PDMS silicones typically behaved significantly stiffer than the organic tissues. Nevertheless, the PDMS silicones developed provided an improved response in comparison to the Silastic 3481 simulant. To practically understand the relative merits of the presented silicone technologies in sports specific applications, human surrogates are required. It is suggested that a multi-material surrogate embodying skin and subcutaneous adipose tissue simulants developed using these techniques potentially offers an improved overall human response and is the subject of further study.

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Figures & Tables

	Part A			Catalyst	Part B		Part A:B Ratio
	Polymers				H301 Cross Linker	V31 Polymer	
	V46	V31	V21				
<i>Relaxed Muscle</i>	46.8%	28.8%	14.4%	10%	10%	90%	4:1
<i>Contracted Muscle</i>	46.8%	21.6%	21.6%				3:1

Table 6 - Part A and B silicone constituent concentrations (% by mass)

		C_I	C_{0I}	$RMS\ Error\ (MPa)$
<i>Relaxed Muscle</i>	0.4s ⁻¹	-4.07×10 ³	8.29×10 ³	0.293
	70s ⁻¹	-4.00×10 ⁻³	1.57×10 ⁴	0.0368
	112s ⁻¹	2.24×10 ⁵	-3.49×10 ⁴	0.0890
<i>Contracted Muscle</i>	0.4s ⁻¹	6.00×10 ⁴	-1.12×10 ³	0.00769
	50s ⁻¹	-7.76×10 ⁴	1.23×10 ⁵	0.0203
	194s ⁻¹	-1.70×10 ⁵	3.03×10 ⁵	0.229

Table 7 - Mooney-Rivlin model coefficients for PDMS simulants

		μ	α	$RMS\ Error\ (MPa)$
<i>Relaxed Muscle</i>	0.4s ⁻¹	1.72×10 ⁴	5.14	0.0232
	70s ⁻¹	2.51×10 ⁴	7.36	0.00715
	112s ⁻¹	3.29×10 ⁵	1.50	0.00942
<i>Contracted Muscle</i>	0.4s ⁻¹	1.16×10 ⁵	1.73	0.00799
	50s ⁻¹	1.66×10 ⁵	7.40	0.0658
	194s ⁻¹	4.02×10 ⁵	7.22	0.730

Table 8 - Ogden model coefficients for PDMS Simulants

		C_{10}	$RMS\ Error\ (MPa)$
<i>Relaxed Muscle</i>	0.4s ⁻¹	1.38×10 ⁴	0.0843
	70s ⁻¹	1.34×10 ⁴	0.0931
	112s ⁻¹	1.67×10 ⁵	0.0995
<i>Contracted Muscle</i>	0.4s ⁻¹	5.81×10 ⁴	0.0119
	50s ⁻¹	8.92×10 ⁴	0.176
	194s ⁻¹	2.15×10 ⁵	1.88

Table 9 - Neo Hooke model coefficients for PDMS simulants

		i	$g(i)$	$\tau(i)$
<i>Relaxed Muscle</i>	1	1.28×10 ⁻¹	3.52×10 ⁻¹	
	2	5.29×10 ⁻²	8.07	
	3	3.39×10 ⁻²	7.61×10 ¹	
<i>Contracted Muscle</i>	1	4.83×10 ⁻²	3.25×10 ⁻¹	
	2	3.66×10 ⁻²	6.91	
	3	2.70×10 ⁻²	8.49×10 ¹	

Table 10 - Prony series coefficients for PDMS simulants

Figure 1 – Schematic showing a typical sports safety standard protocol used to assess projectile impacts (BS:13546:2002 Field Hockey Goalkeeper Protective Clothing; BS:6183-3:2000 Cricket Leg Protectors)

Figure 17 – Compressive engineering stress-strain plot showing data from previous organic tissue characterisation studies

Figure 18 – Engineering stress-strain graphs showing organic tissue properties from relaxed (McElhaney, 1966; Song *et al.*, 2007) and contracted (Zheng *et al.*, 1999) muscle tissue in (a) linear and (b) log-linear plots

Figure 19 – Quasi-static compressive engineering stress-strain plot of silicones with different cross-linker concentrations in: (a) linear and (b) log-linear graphs

Figure 20 - Chemical structure of additive cure PDMS silicone reaction

Figure 21 - Programmed triangular loading profile for quasi-static uniaxial compressive tests

Figure 22 - Processing of intermediate strain rate drop tower data: (a) raw data; (b) smoothed stress-strain plot

Figure 23 – Quasi-static stress-strain plots showing the divergences in compressive response between: (a) PDMS relaxed muscle simulant and organic tissue data; (b) PDMS contracted muscle simulant with organic tissue data; (c) Silastic 3481 with relaxed organic tissue data; (d) Silastic 3481 with contracted muscle organic tissue data; (e) Percentage errors plots showing divergence between silicone simulants and target organic contracted muscle data

Figure 24 - Engineering stress-strain plot showing stress softening effect in cyclic compressive protocol between the first and last cycles in: (a) PDMS relaxed muscle simulant and (b) PDMS contracted muscle simulant

Figure 25 – Intermediate strain rate engineering stress-strain curves for: (a) PDMS relaxed muscle simulant and (b) PDMS contracted muscle simulant

Figure 26 Normalised stress-time plot showing stress relaxation in PDMS relaxed and contracted muscle simulants

Figure 27 – Log-linear engineering stress-strain plots showing a comparison between Silastic 3481, PDMS silicones and organic tissue properties for: (a) PDMS relaxed muscle simulant and (b) PDMS contracted muscle simulant

Figure 28 – Hyperelastic model fits for: (a) PDMS relaxed muscle simulant and (b) PDMS contracted muscle simulant at a range of intermediate strain rates

Figure 29 - Prony series model fits for: (a) PDMS relaxed muscle simulant and (b) PDMS contracted muscle simulants

Figure 30 – Intermediate strain rate PDMS relaxed muscle simulant responses compared with human muscle characterisation studies by Chawla et al. (2009) and Balaraman et al. (2012)

Figure 31 - Normalised stress-time plots showing differences in stress relaxation between organic muscle tissue (Van Looke et al., 2008) and the PDMS relaxed and contracted muscle simulants