# Mechanical Response of the Human Sub-Calcaneal Heel Pad under High Rate Compression

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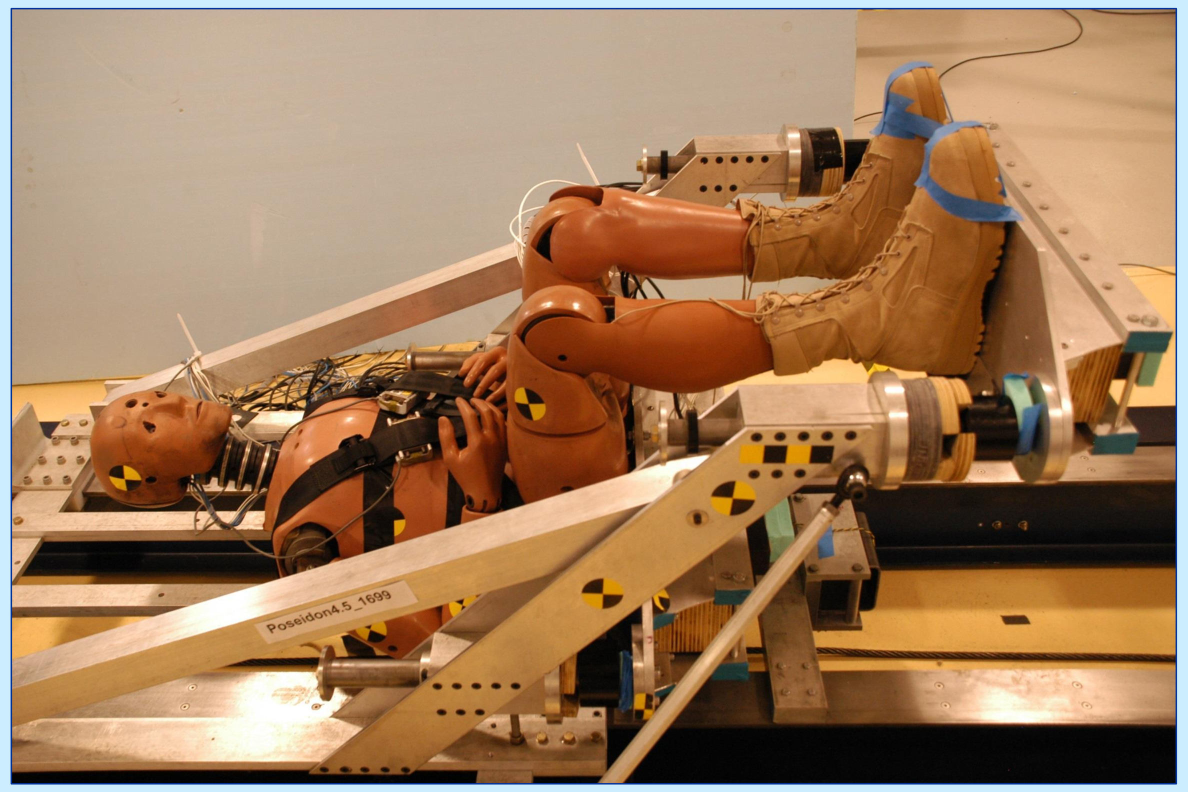
## Introduction

- In the instance of a high rate axial load to the lower limb as a result of an under-vehicle blast, the soft tissue layer in the plantar region of the foot is the first structure engaged.
- A material characterization of this structure under such loading conditions would provide a better understanding of the load paths to the lower extremity, and accurate material properties for the development of biofidelic anthropometric test devices (ATDs).

### Goal

- Given that the current Hybrid-III ATD is known to lack accurate heel compression characteristics for high rate loading, it is the goal of this study to acquire accurate material properties of the human sub-calcaneal heel pad under high rate compressive loading.

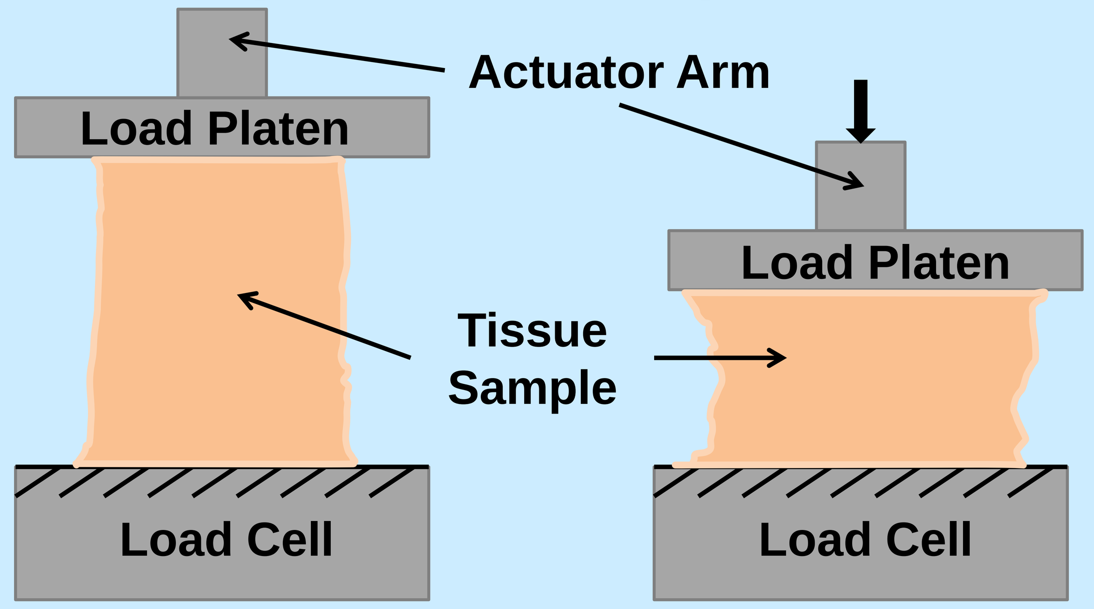


## Methods

- Protocols for the handling of biological materials were approved by the University of Virginia's Institutional Biosafety Committee.
- Three heel pads were collected from the hind foot region of 2 post mortem human surrogates.
- Heel pads were flash frozen and stored in a morgue freezer at -20°C until materials testing could be performed.
- Two to four tissue samples were cut to approximately 10mm in diameter and 10mm in height from each whole heel pad using a cylindrical boring tool. For ease of cutting, the heel pads were left partially frozen while being prepared.
- Samples were cut perpendicular to the surface of the skin and from the central portion of the pad.
- The quality of each sample was evaluated and those that were not of good cylindrical shape were discarded.
- Samples underwent a battery of ramp and hold stress relaxation tests for material characterization.

## Test Conditions

- Materials testing was performed on a bench-top test machine (ElectroForce® 3100, Bose, Eden Prairie, MN).
- Samples were placed on an aluminum stage mounted atop a 250gram capacity Load Cell, and beneath a flat aluminum load platen mounted to a linear actuator equipped with an LVDT to measure displacement.
- Force and displacement data were acquired at 20kHz.




- Samples were compressed up to 20% engineering strain at peak rates between (150-350)mm/s, (15-35)$s^{-1}$ with a test duration of 30s.
- A 2gram pre-compression was placed on the samples to ensure contact.
- Data were filtered using a zero-phase, digital IIR 8 pole Butterworth filter at a Low Pass frequency of 1650Hz and resampled in a logarithmically scaled time step.
- Tissue stresses were calculated by dividing the force data by the un-deformed, initial cross-sectional area. The initial unstrained height of the sample was measured as the distance between the platens and used to calculate tissue strain.

## Mathematical Formulation

- The stress output $\sigma(\varepsilon, t)$, due to the strain input $\varepsilon = \varepsilon(t)$, was modeled using quasilinear viscoelasticity (QLV).

$$\sigma(\varepsilon, t) = \int_0^t G(t - t') \frac{\partial \sigma^e(\varepsilon)}{\partial \varepsilon} \frac{\partial \varepsilon}{\partial t'} dt'$$

- where $\sigma^e(\varepsilon)$ is the **instantaneous elastic response** and $G(t)$ is the **reduced relaxation function**. Two nonlinear constitutive models were considered for $\sigma^e(\varepsilon)$: First, the tissue was modeled as a 1D material.

$$\sigma^e(\varepsilon) = a_1\varepsilon + a_2\varepsilon^2 + a_3\varepsilon^3$$

- where $a_i$ are constants. Second, the tissue was assumed to be incompressible and isotropic and modeled using an exponential strain energy density function (SEDF).

$$W = \frac{\mu_o}{2\gamma}\left[e^{\gamma(I_1-3)} - 1\right]$$

- where $\mu_o$ is the elastic shear modulus, $\gamma$ is the nonlinearity coefficient, and $I_1$ is the first invariant of the right Cauchy-Green strain tensor. Assuming uniaxial compression, the 2nd PK stress in the direction of loading is derived as

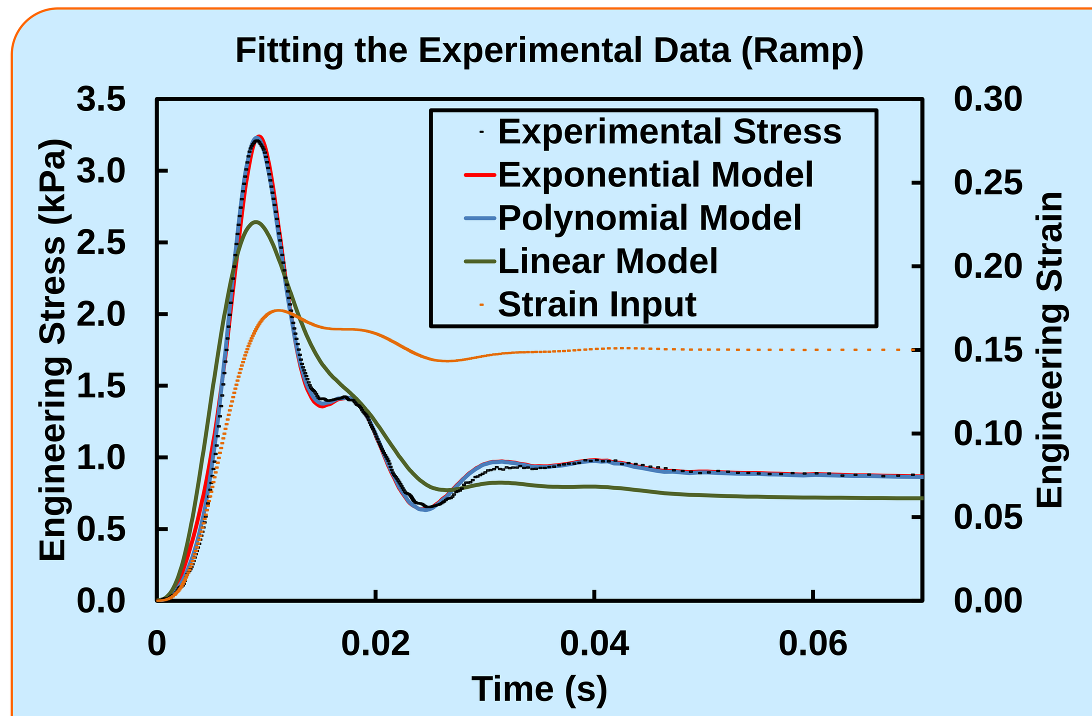
$$\sigma^e(\lambda) = \frac{\mu_o e^{\left[\gamma\left(\lambda^2 + \frac{2}{\lambda} - 3\right)\right]}(\lambda^3 - 1)}{\lambda^3}$$

- where $\lambda = \varepsilon + 1$ is the stretch ratio. $G(t)$ was chosen to be

$$G(t) = G_\infty + \sum_{i=1}^{5} G_i \cdot e^{-\frac{t}{\tau_i}}$$

- where $G_i$'s are the normalized relaxation coefficients and $\tau_i$'s are the time constants, $G_\infty + G_1 + G_2 + G_3 + G_4 + G_5 = 1$, and $\tau_1 = 0.001$s, $\tau_2 = 0.01$s, $\tau_3 = 0.1$s, $\tau_4 = 1$s, $\tau_5 = 10$s.

## Results



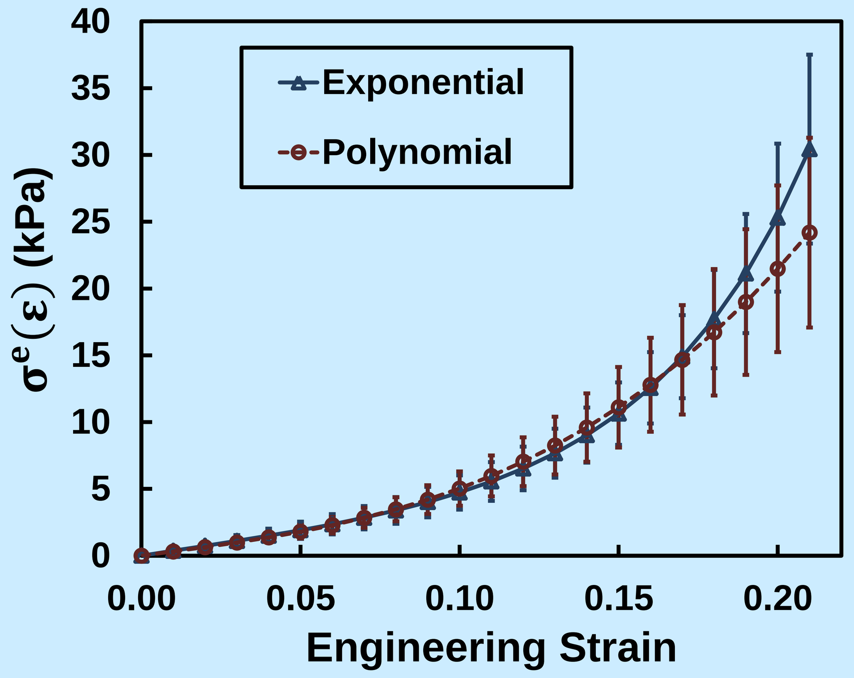

- Seventeen total compression tests were performed.
- Values for the coefficients of $\sigma^e(\varepsilon)$ and $G(t)$ were determined using a reduced gradient algorithm (Excel Solver®, Microsoft®, Redmond, WA) to minimize the sum squared error between the model-predicted force and experimental data.
- An average $\sigma^e(\varepsilon)$ and $G(t)$ were determined through the previously described reduced gradient algorithm for both constitutive models.
- The instantaneous elastic shear modulus was determined to be **(12.6 ± 3.7)kPa**.

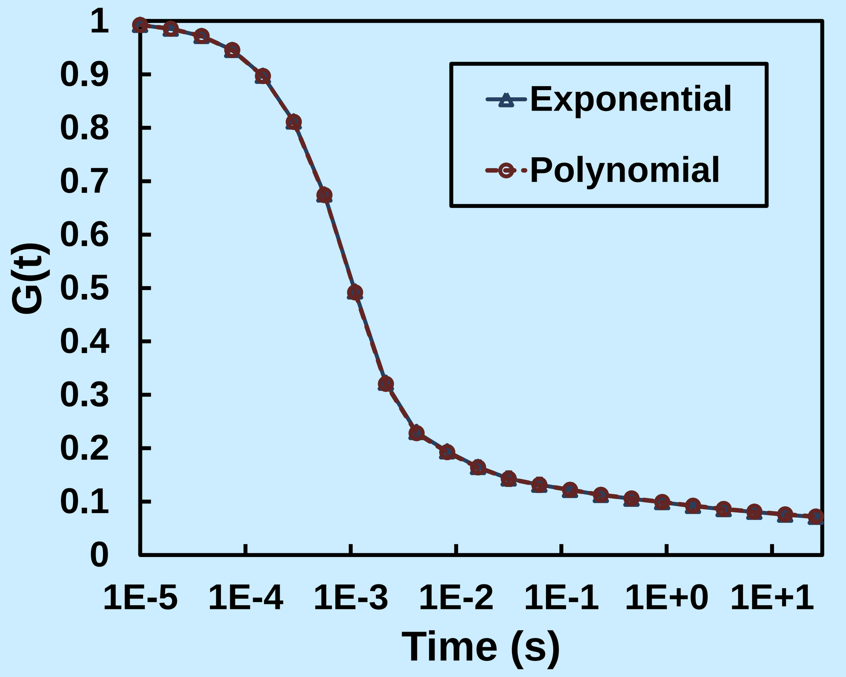
**Exponential**

| Unit | Coef | Value ± 95%CI |
|---|---|---|
| kPa | $\mu$ | 12.6 ± 3.7 |
| - | $\gamma$ | 14.7 ± 2.3 |
| - | $G_1$ | 0.741 ± 0.007 |
| - | $G_2$ | 0.110 ± 0.010 |
| - | $G_3$ | 0.034 ± 0.004 |
| - | $G_4$ | 0.023 ± 0.002 |
| - | $G_5$ | 0.022 ± 0.001 |
| - | $G_\infty$ | 0.070 ± 0.010 |

**Polynomial**

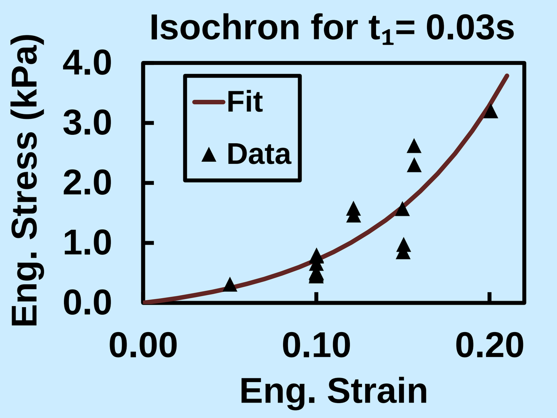
| Unit | Coef | Value ± 95%CI |
|---|---|---|
| kPa | $a_1$ | 31.2 ± 7.8 |
| kPa | $a_2$ | 0.00 ± 0.0 |
| kPa | $a_3$ | 1903 ± 624 |
| - | $G_1$ | 0.744 ± 0.009 |
| - | $G_2$ | 0.108 ± 0.015 |
| - | $G_3$ | 0.033 ± 0.005 |
| - | $G_4$ | 0.023 ± 0.002 |
| - | $G_5$ | 0.021 ± 0.001 |
| - | $G_\infty$ | 0.070 ± 0.010 |

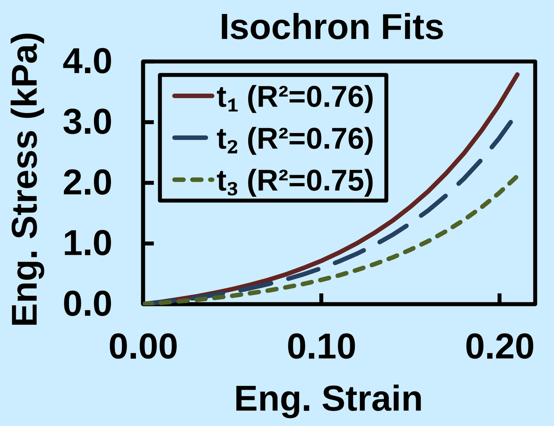



## Discussion

- The tissue behaved viscoelastic and spatially nonlinear.
- At strains above 17% the $\sigma^e(\varepsilon)$ curves began to diverge and the exponential model predicted higher stresses. For strains below 17% the models were approximately equal.
- In sixteen out of seventeen tests, the polynomial model resulted in higher $R^2$ values and lower Sum Squared Error indicating a better fit.
- $G(t)$ was nearly identical for both models.
- The shear modulus is in agreement with values reported in the literature; (8-16)kPa.
- A linear viscoelastic model was fit to the data, in addition to both QLV models, but did not capture the ramp and peak stress.
- The assumption of QLV was justified using the method of isochrones. Three isochrones were chosen: $t_1$= 0.03s, $t_2$= 0.10s, and $t_3$= 5s after the peak stress. Stress vs. strain data for each isochron were plotted and fit with an exponential function $\alpha\left(e^{\beta\varepsilon} - 1\right)$, which provided a better fit over a linear function. Dividing the isochronous curves resulted in approximately constant values indicating no temporal nonlinearity up to 20% strain.




### Limitations

1. The tissue was mechanically damaged through preparation protocols which may have significantly altered the material properties.
2. Ensuring a flat loading surface and uniform, unconfined deformation of the tissue under compressive loading.

## Conclusions

- Material properties of the sub-calcaneal heel pad were determined for engineering strains up to 20% and strain rates between (15-35)$s^{-1}$.
- QLV was validated using the method of isochrones and used to fit two constitutive models to the experimental data.
- The polynomial model indicated a better fit to the data.
- The shear modulus from the exponential SEDF agrees with the values reported in the literature.

### Future Research

1. Data will be collected up to 50% engineering strain.
2. Heel pad from the current Hybrid-III ATD will be characterized and the material properties compared to those determined for humans.
3. Material properties will be validated through finite element analyses.

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