DENSITY CHANGES IN BOVINE TENDON RESULTING FROM BUFFERED AND UNBUFFERED SOLUTIONS

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INTRODUCTION

Standard engineering materials can be characterized with a variety of parameters including elastic modulus, yield stress, and ultimate tensile strength. Many engineering materials behave in a well-characterized fashion, however, when applying the same methods to biological materials their performance is varied, and highly dependent upon the methods of *in-vitro* preparation. This has led to a variety of complications in biological materials research. The objective of this work was to address the issue of cross-sectional area prediction of biological tissues. One method, in particular, uses mass and length measurements and assumed values for tissue density to determine an average cross-sectional area as per the formula [1].

$$\frac{m}{dL} = \overline{A} \quad (1)$$

Where
d is the density of the specimen
m is the mass of the specimen
L is the length of the specimen and
$\overline{A}$ is the average cross-sectional area.

Using the definition of engineering stress, there is an inverse relationship between stress and average cross-sectional area.

$$\sigma = \frac{F}{\overline{A}} = F \times \frac{dL}{m} \quad (2)$$

Where
F is the force applied and
σ is stress.

Tissue volume can be determined through fluid displacement. In itself this is not a problem; however, the submersion does bring into question how the tissue dimensions and density change in *in-vitro* solutions. Unbuffered hyper- or hypo-tonic solutions may cause bulk fluid transfer, changing the density of the sample as well as the overall sample dimensions [1]. Because submersion affects fluid retention, it is difficult to determine pre- and post-volume measurements. However, length and mass measurements may be performed independent of submersion with changes in this dimension being used to determine the sensitivity of a tissue to submersion. In this manner, the effect on the change in predicted stress ($\Delta\sigma$) may be approximated.

$$\therefore \Delta\sigma \cong \left(\frac{FdL_0}{m_o}\right) - \left(\frac{Fd\left(L_0 + \delta_{Length}\right)}{(m_o + \delta_{mass})}\right) \quad (3)$$

Where
δ_{Length} is the change in sample length
δ_{mass} is the change in sample mass

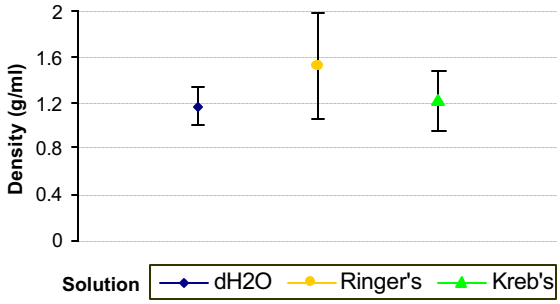

Figure 1: Predicted tissue densities for a 60 second submersion (mean±standard deviation).

METHODS

Three solutions were prepared (Ringer's Solution, Kreb's Solution and deionized water (dH_2O)). Bovine hoof tendon samples (n=14) were obtained from the Ontario Veterinary College within one hour of slaughter and dissected longitudinally into 80mm lengths. Sample lengths and masses were evaluated pre- and post- submersion. All length measurements were made in triplicate using a MicroScribe™ 3D Digitizer (Immersion Corporation, San Jose, CA, USA). Tendon samples from each cow were submerged in one of the three solutions for approximately 60 seconds and their volumes were determined by their fluid displacement. A system of syringes was created for this purpose. Density was determined from the volume and the post-mass measurements.

RESULTS AND DISCUSSION

No significant post- minus pre-immersion changes in mass ($p=0.73$) or sample length ($p=0.15$) were observed as a result of solution type. However, a significant solution effect was found in the density values ($p=0.01$) where the Ringer's solution tissue had higher values for density than the dH_2O or the Kreb's solutions. Interestingly there was no difference observed for stress variability ($\Delta\sigma$) by solution ($p=0.89$) indicating that solution type did not alter the predicted stress.

CONCLUSIONS

Results suggest that while solution type affects density, any effects on predicted stress cancel out probably as a result of subtle mass and length changes in the tissue sample.

REFERENCES

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