Published in final edited form as:

Comput Methods Biomech Biomed Engin. 2017 May ; 20(6): 598–601. doi:
10.1080/10255842.2016.1268132.

A case for poroelasticity in skeletal muscle finite element analysis: experiment and modeling

Benjamin B Wheatley,

Soft Tissue Mechanics Laboratory, Department of Mechanical Engineering, Colorado State University, 1374 Campus Delivery, Fort Collins, CO 80523, (970) 297-4033, wheatbenleyjamin@gmail.com

Gregory M Odegard,

Department of Mechanical Engineering – Engineering Mechanics, Michigan Technological University, 1400 Townsend Drive, Houghton, MI 4993, (906) 487-2329, gmodegar@mtu.edu

Kenton R Kaufman, and

Motion Analysis Laboratory, Department of Orthopedic Surgery, Mayo Clinic, 200 First Street SW, Rochester, MN 55906, (507) 284-2262, kaufman.kenton@mayo.edu

Tammy L Haut Donahue

Soft Tissue Mechanics Laboratory, Department of Mechanical Engineering, School of Biomedical Engineering, Colorado State University, 1374 Campus Delivery, Fort Collins, CO, 80523, (970) 491-1319, Tammy.Donahue@colostate.edu

Abstract

Finite element models of skeletal muscle typically ignore the biphasic nature of the tissue, associating any time dependence with a viscoelastic formulation. In this study, direct experimental measurement of permeability was conducted as a function of specimen orientation and strain. A finite element model was developed to identify how various permeability formulations affect compressive response of the tissue. Experimental and modeling results suggest the assumption of a constant, isotropic permeability is appropriate. A viscoelastic only model differed considerably from a visco-poroelastic model, suggesting the latter is more appropriate for compressive studies.

Keywords

biphasic; permeability; transversely isotropic; viscoelasticity

Introduction

In addition to muscle fibers and extracellular matrix, skeletal muscle consists of approximately 75% fluid (Sjøgaard and Saltin, 1982). Thus, it can be characterized as a biphasic material, consisting of both a solid and a fluid phase. Other biphasic biological tissues such as cartilage (Mansour and Mow, 1976) and meniscus (LeRoux and Setton,

2002) utilize fluid content to distribute under loads compression. However, the biphasic properties of muscle, specifically permeability, are not known. The role which permeability plays in skeletal muscle mechanics, known as poroelasticity, is unclear (Yang and Taber, 1991). In computational models of skeletal muscle, time dependence is often neglected (Van Looke et al., 2006) or approximated using viscoelastic modeling (Van Looke et al., 2008; Wheatley et al., 2015). The goals of this work were to 1) characterize skeletal muscle permeability by direct experimental measurement, and 2) identify how various permeability implementations affect skeletal muscle models in compression.

Methods

Four New Zealand White Rabbits ~2.5 kg in weight, were euthanized with Colorado State University Institutional Animal Care and Use Committee approval. One biceps femoris muscle from each animal was harvested from the left or right hind limb. All procedures were completed within 6 hours post mortem to reduce the effects of rigor mortis (Van Looke et al., 2006). Cylindrical samples (4 mm height and 8 mm diameter) were removed from the muscle mid-belly using a drop cutter and a biopsy punch. As skeletal muscle exhibits transversely isotropic passive behavior (Pietsch et al., 2014; Takaza et al., 2012), samples were obtained in the longitudinal and the transverse directions.

The permeability of each sample was directly evaluated using a permeation test device similar to previously published work (Weiss and Maakestad, 2006). In short, this device applied a known flow rate (0.5 mL/min) with a syringe pump to phosphate buffered saline across a tissue sample while measuring the pressure difference across the sample (Figure 1A). Additionally, since permeability is typically a function of tissue strain (Weiss and Maakestad, 2006), samples were compressed axially (12.5% and 25%) to replicate transverse or longitudinal compression or compressed laterally (25% and 50%) to simulate longitudinal stretch (Figure 1B–D). Pressure (P), flow rate (Q), specimen dimensions (L_0 and A), and strain (ϵ) were measured to calculate the permeability of the tissue (Weiss and Maakestad, 2006) (Equation 1). A paired t-test was utilized to compare permeability values between longitudinal and transverse samples and a general linear model was utilized to determine if permeability varied with strain level ($p < 0.05$ for both analyses).

$$k = \frac{QL_0(1+\epsilon)}{AP} \quad (1)$$

A 3D finite element (FE) model representing midbelly muscle tissue was developed using FEBio (febio.org) with 270 hexahedron elements. This model simulated a 4×4×20 mm cuboid using two planes of symmetry and had muscle fibers aligned parallel to the length of the cuboid. The cuboid was compressed in the transverse direction under stress relaxation (20% strain at 10% second⁻¹ followed by a 300 second relaxation) before compression to 50% strain at 1% second⁻¹. A poroelastic ellipsoidal fiber reinforced coupled (compressible) Mooney-Rivlin constitutive model was implemented. Strain-dependent permeability was implemented as a function of the volume ratio J , the undeformed void ratio ϕ_0 (0.19 for

skeletal muscle (Sjøgaard and Saltin, 1982)), and material properties k_0 , M , and α (Equation 2) (Weiss and Maakestad, 2006).

$$k(J) = k_0 \left(\frac{J - \varphi_0}{1 - \varphi_0} \right)^\alpha \exp \left[\frac{M}{2} (J^2 - 1) \right] \quad (2)$$

To study how various permeability assumptions affects model behavior, four separate formulations were utilized in separate simulations: a solid analysis only as a baseline (“Solid”), a constant, isotropic permeability for a simple biphasic model (“Constant”), and two separate fits of Equation 2 to experimental data to generate anisotropic and strain dependent responses (“Fit I” and “Fit II”) (Table 1). Fit I was a conservative strain dependent formulation while Fit II was a case of highly nonlinear strain dependence similar to other tissues (Weiss and Maakestad, 2006). Additionally, the “Constant” model was compared against two models incorporating viscoelasticity of the solid phase through a Prony series (Wheatley et al., 2015), one with both poroelasticity and viscoelasticity, and one with viscoelasticity only. The mean Cauchy stress in the direction of compression was compared between models.

Results

The mean permeability value for all samples was $7.41 \times 10^{-11} \text{ m}^4/\text{N}\cdot\text{s}$ with a standard error of $2.2 \times 10^{-11} \text{ m}^4/\text{N}\cdot\text{s}$. There was no statistical effect of either strain level ($p=0.398$) or orientation ($p=0.158$) on skeletal muscle permeability (Figure 2A–B).

Utilizing a biphasic approach affected tissue behavior particularly under transient conditions (Figure 2C–D and Table 2), while variations in strain dependence and anisotropy played less of a role than poroelasticity itself. A combined visco-poroelastic model differed drastically from all other analyses, including the viscoelastic only model (Figure 2E–F and Table 2).

Discussion

This work provides, to the authors’ best knowledge, the first direct measurement of skeletal muscle permeability. The observed values from this work show an increase of 4–5 orders of magnitude over other biological tissues such as cartilage or ligament (Mansour and Mow, 1976; Weiss and Maakestad, 2006). This is not surprising since low muscle permeability would greatly resist muscle deformation, thus reducing contractile capabilities.

Model calculated stresses had good agreement with previously published experimental work under compression (Pietsch et al., 2014; Van Looke et al., 2006) which gives greater confidence in these results. Statistical results showed no dependence of permeability on strain level or orientation, while modeling results indicated strain and orientation dependence played less of a role than even a simple poroelastic model versus a solid model. Additionally, the viscoelastic comparison data suggest viscoelasticity alone may not always be appropriate for skeletal muscle modeling in compression. Thus, a simple biphasic analysis (isotropic, constant permeability) combined with viscoelasticity could be

appropriately utilized for skeletal muscle models involving highly transient compressive conditions such as impact.

Acknowledgments

This work was funded by the National Institutes of Health: National Institute of Child Health & Human Development [R01HD31476-12].

References

1. LeRoux MA, Setton LA. Experimental and Biphasic FEM Determinations of the Material Properties and Hydraulic Permeability of the Meniscus in Tension. *J. Biomech. Eng.* 2002; 124:315. [PubMed: 12071267]
2. Mansour JM, Mow VC. The permeability of articular cartilage under compressive strain and at high pressures. *J. Bone Joint Surg. Am.* 1976; 58:509–16. [PubMed: 1270471]
3. Pietsch R, Wheatley BB, Haut Donahue TL, Gilbrech R, Prabhu R, Liao J, Williams LN. Anisotropic compressive properties of passive porcine muscle tissue. *J. Biomech. Eng.* 2014; 136:111003.
4. Sjøgaard G, Saltin B. Extra- and intracellular water spaces in muscles of man at rest and with dynamic exercise. *Am. J. Physiol.* 1982; 243:R271–80. [PubMed: 7114288]
5. Takaza M, Moerman KM, Gindre J, Lyons G, Simms CK. The anisotropic mechanical behaviour of passive skeletal muscle tissue subjected to large tensile strain. *J. Mech. Behav. Biomed. Mater.* 2012; 17:209–220. [PubMed: 23127635]
6. Van Loocke M, Lyons CG, Simms CK. A validated model of passive muscle in compression. *J. Biomech.* 2006; 39:2999–3009. [PubMed: 16313914]
7. Van Loocke M, Lyons CG, Simms CK. Viscoelastic properties of passive skeletal muscle in compression: Stress-relaxation behaviour and constitutive modelling. *J. Biomech.* 2008; 41:1555–1566. [PubMed: 18396290]
8. Weiss, Ja, Maakestad, BJ. Permeability of human medial collateral ligament in compression transverse to the collagen fiber direction. *J. Biomech.* 2006; 39:276–83. [PubMed: 16321629]
9. Wheatley BB, Pietsch RB, Haut Donahue TL, Williams LN. Fully non-linear hyper-viscoelastic modeling of skeletal muscle in compression. *Comput. Methods Biomech. Biomed. Engin.* 2015:1–9.
10. Yang M, Taber LA. The possible role of poroelasticity in the apparent viscoelastic behavior of passive cardiac muscle. *J. Biomech.* 1991; 24:587–97. [PubMed: 1880142]

Figure 1.

A) Permeability testing apparatus, where fluid (blue) at a pressurized level P passes through the specimen (red) at a flow rate Q , across an area A , and along a length $L = L_0(1 + \epsilon)$. B–D) Specimens were compressed axially to simulate B) transverse and C) longitudinal compression or laterally to simulate D) longitudinal tension.

Figure 2.

Experimental permeability raw data ($\pm$ standard error of the mean), Fit I, and Fit II curves for A) longitudinal samples and B) transverse samples. Cauchy stress outputs for the various biphasic analyses for C) initial 20% compression ramp and relaxation and D) 45–50% compression and for the viscoelastic comparisons for E) initial 20% compression ramp and relaxation and F) 45–50% compression.

Table 1

Parameter values for the three utilized permeability formulations.

Permeability Model	k_0 (m ⁴ /N-s)		M (-)		α (-)	
	Long	Trans	Long	Trans	Long	Trans
Isotropic Constant	7.41		N/A	N/A	N/A	N/A
Fit I	11	4.2	0.63	0.16	0.1	0.97
Fit II	50	5	10	1	2	1.5

Table 2
Mean Cauchy stress values for all analyses at compressive strains of 20% (peak response) and 50%.

Mean Cauchy Stress in kPa	Solid	Constant	Fit I	Fit II	Visco Only	Visco-Poro
20% Strain Peak	-1.01	-1.52	-1.71	-1.66	-1.74	-2.32
50% Strain	-5.66	-5.91	-5.96	-6.10	-6.95	-7.19