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A review of the characterizations of soft tissues used in human body modeling: Scope, limitations, and the path forward

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ABSTRACT

Soft tissue material properties are vital to human body models that evaluate interactions between the human body and its environment. Such models evaluate internal stress/strain responses in soft tissues to investigate issues like pressure injuries. Numerous constitutive models and parameters have been used to represent mechanical behavior of soft tissues in biomechanical models under quasi-static loading. However, researchers reported that generic material properties cannot accurately represent specific target populations due to large inter-individual variability. Two challenges that exist are experimental mechanical characterization and constitutive modeling of biological soft tissues and personalization of constitutive parameters using non-invasive, non-destructive bedside testing methods. It is imperative to understand the scope and appropriate applications for reported material properties. Thus, the goal of this paper was to compile studies from which soft tissue material properties were obtained and categorize them by source of tissue samples, methods used to quantify deformation, and material models used to describe tissues. The collected studies displayed wide ranges of material properties, and factors that affected the properties included whether tissue samples were *in vivo* or *ex vivo*, from humans or animals, the body region tested, body position during *in vivo* studies, deformation measurements, and material models used to describe tissues. Because of the factors that affected reported material properties, it is clear that much progress has been made in understanding soft tissue responses to loading, yet there is a need to broaden the scope of reported soft tissue material properties and better match reported properties to appropriate human body models.

1. *Introduction*

Appropriate soft tissue material properties are integral to modeling the human body, particularly during interactions with objects in its environment (e.g. chairs, assistive devices, etc.). Finite element (FE) models are important tools that describe internal stresses and strains in the body's soft tissues resulting from such interactions and have been used to evaluate automotive seating comfort, office seating, and the risk of pressure injuries (PIs) for wheelchair users [1–4]. Models that predict PI risk are particularly good examples of the need for appropriate material properties, specifically because internal stresses and strains have been related to PI formation and were shown to be affected by marginal changes in tissue stiffness [5]. Several researchers investigated the sensitivity of models to differences in material properties, showing that differences in material properties within one order of magnitude, particularly the muscle material properties, changed tissues stresses and strains by as much as 70% [6,7]. Furthermore, such models that evaluate stresses and strains should use material properties from populations of elderly people and wheelchair using individuals, since they are at the highest risk and also experience significant changes in their tissue composition [8,9]. However, recent models have used soft tissue properties of animals for two main reasons, 1) there is a dearth of material properties representing human tissue and 2) in general, testing with animals is easier than testing with humans for material property collection [10,11]. These challenges are compounded ever further if one wishes to obtain personalized, human material property data in the clinic, or at the bedside. Though there is a clear need for material property data for use in models that evaluate risk of PI formation, the need for appropriate material properties extends to all human body models.

Ideally, each human body model would use personalized data in the model, however there have been numerous barriers to doing so. Historically, it has been easier for researchers to obtain data from animals than humans. If human data were collected, it also was most typically in

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a laboratory setting. However, recently, some researchers conducted studies to determine soft tissue properties and geometries in clinical settings [12–15]. Of these studies, two of the four used magnetic resonance imaging (MRI) or ultrasound (US), both of which were available in clinical settings, whereas the third sought to develop a relatively affordable tool to determine anatomical geometry without the use of MRI or US, and the last determined material properties without imaging. One of the more common methods for determining soft tissue properties has been shear wave elastography, which was able to use MRI or US images provided that the proper loading data were collected as well [16, 17]. A commonality between these studies was that the results were obtainable in a relatively short time frame (under half an hour), which made them possible to implement during a scheduled clinical visit. Yet, while there have been efforts to obtain personalized, in-clinic soft tissue property, these situations were very limited. Therefore, a need still exists for new measurement approaches to quickly and easily obtain data sets from a broad range of body regions for use in soft tissue material property determination.

Several factors have been shown to affect material properties used in models, namely, the tissue being *in vivo* or *ex vivo*, whether the tissue properties were from humans or animals, the body region of the properties, and the body position. Material properties of tissues have been obtained from a variety of sources, including *in vivo* and *ex vivo* animal and human tissues. Though it was noted as a limitation, FE models still used tissue properties from animal tissue (rats, sheep, pigs, and cows) when appropriate data from humans were not available [5,18–22]. Limited data from *in vivo* human tissue have been collected from multiple body regions in humans in different body positions; whereas both body region and position were found to affect material properties [23–25]. Further, the inclusion or exclusion of persons with physical disabilities, such as persons with spinal cord injuries, has been shown to affect the material properties, as tissue compositions from the individuals with spinal cord injury differed from those of able-bodied individuals [9,26–35]. Considering factors such as the source of tissues properties from animals or humans, whether the tissues were *in vivo* or *ex vivo,* population, body position, and anatomical region is crucial to the accuracy of the FE models.

Another challenge with the application of material properties is consideration of the experimental setup used to collect the material property data, as several types of experimental methods were used to collect load and deformation data from tissue samples. Load data have been collected in the form of force during indentation of specific regions of tissue or by recording pressures on support surfaces that contact tissue [36,37]. Deformation was defined several ways, including displacement, strain, or stretch in the tissue, and several methods have been used to obtain the deformation measurements of the tissues [34,38–51]. These methods included tensile testing, suction testing, indentation testing, compression testing, shear wave elastography, ultrasound, magnetic resonance elastography, and digital image correlation. The numerous methods used to collect load and deformation data may confound reported properties as well.

Lastly, several material models were used to describe the mechanical behavior of soft tissues. Early experiments described tissues as linear elastic materials, however more recent investigations characterized tissue as either non-linear materials (e.g. Neo Hookean, Fung) or viscoelastic materials [43,47,52]. The choice of material model determined the resulting material parameters, which were specific to each model [53–55]. The models typically used to describe each type of tissue (muscle, fat, and skin) were not the same, meaning the reported parameters were different, and thus difficult to compare to one another. Further complicating comparisons between existing characterizations of soft tissues was the lack of a consistently defined stress-free, or reference, state. This is important because it could not be assumed that tissues *in vivo* or *ex vivo* were stress-free; and as such, tissues in past studies likely had differing pre-strain histories, affecting results [56,57]. Though there is ongoing research with the goal of defining stress-free states for soft tissues, a consistent method of determining stress-free states has not been used across studies [58–61]. Since many of the models used to describe soft tissue were non-linear or viscoelastic, their behavior was dependent on residual stresses and strains in the tissues, which were not comparable between studies. A method has yet to be developed to convert between sets of material parameters designated for each material model without the original load and deformation data, even before considering pre-strain histories. This means that, although there are preferred material models for each tissue, there is no standard material characterization for each tissue.

Numerous studies have been conducted to quantify the material properties of soft tissues over the past few decades. However, many human body model creators acknowledged that the material properties they used for their models may not have been appropriate for the target population of their study [11,62–64]. Because there are ranges of material properties for each soft tissue, depending on the experimental conditions in which they were determined, numerous researchers have called for, and provided, guidelines for their usage in any biomechanical model [65–70]. Among these guidelines are sensitivity, verification, and validation studies to quantify the effects of using marginally different material properties on the outputs of models. These studies allow for easier communication of models between researchers by explaining, identifying, and quantifying areas of uncertainty. This is especially true when explaining if and how differences between model predictions of measurable outputs, and experimental measurements of those outputs themselves, occur. Because of all the factors that have been shown to affect reported soft tissue material properties, and how those properties affected model outputs when implemented, a detailed documentation the material properties used and how they were obtained is needed. By documenting the tissue properties that have been reported and under what experimental circumstances (i.e. from what body region, *in vivo* or *ex vivo*, from humans or animals, etc.), it will be possible to evaluate what tissues properties will need to be studied to be implemented in forthcoming human body models. The properties reported can also serve as a benchmark against which future personalized soft tissue property results may be compared Thus, the goal of this paper was to compile the studies from which material properties have been obtained and categorize them by the sources of their tissue samples (*in vivo* human, *ex vivo* human or animal tissue), the methods used to quantify deformation during experiments, and the material models used to describe the tissues. This collection of methods and data will be an invaluable tool for those who determine and use the material properties of soft tissues.

2. Methods

The authors searched for studies reporting material properties for soft tissues of humans and animals. A literature search was conducted from multiple sources: PubMed, Google Scholar, Science Direct, Elsevier, Taylor and Francis Group, Scientific Research, Springer Link, and the Wiley Online Library. The following keywords were used to search for such topics: mechanical properties of human tissue, tissue parameters, material parameters, *in vivo* tissue properties of humans, *in vivo* tissue properties of animals, *ex vivo* tissue properties of humans, and *ex vivo* tissue properties of animals. 76 papers were included in this review. The references of these publications were further analyzed for additional articles to add to the initial search results.

3. Results

The results from this literature review are represented in Tables 1–6. Studies were placed into tables by whether they included *ex vivo human* or *in vivo human* tissue samples or animal samples. They were then *further* categorized by the type of soft tissue in Tables 1–4 and ordered by the chronology of their reference in this review. Each entry of a reference includes the type of tissue investigated, the location on the body from which the sample was taken, the sample size, the testing position

(if applicable), the material parameters found, the model used to describe the tissue, the method used to take deformation measurements, and the location of the equation used to determine the parameters in either Table 5 or 6. To find the equation used to determine the parameters for each entry, the number before the period in the penultimate column gives the table number to use, and the number after the period gives the equation number (i.e. 5.12 is Table 5, equation number 12).

3.1. *Ex vivo human tissue properties*

15 studies were found that reported on *ex vivo* human muscle, fat, and skin tissues, all of which included a Young's modulus, shear modulus, ultimate stress or strength of the tissue, and/or isotropic material constants [24,71–84]. These studies are listed in Table 1.

Table 1
Ex vivo human material properties. Each column provides information about the tissue type, sample location, sample size, parameters found, the model used, mechanical test (deformation definition in parentheses), where the foundational equation can be found in Tables 5 and 6, and the reference.

Tissue Type	Sample Location	Sample Size	Parameters	Constitutive Law	Mechanical Test	Table 5/6	Ref.
Muscle	Leg	N=70 N=13(m) N=11(f)	E = 20-33 kPa (m) E = 12-38 kPa (f)	Linear elastic	Compression test (not reported)	5.1	[71]
	Levator ani	N=9	$k = 4.69x10^3 \pm 2.91x10^3$ kPa $q = 0.44 \pm 0.08$	Power-law	Tensile test (stretch and Green-St. Venant strain)	6.10	[72]
	Multifidus	N=23	$E_{single\ fiber} = 33.7 \pm 1.9$ kPa $E_{fiber\ bundle} = 91.3 \pm 6.9$ kPa	Linear elastic	Microindentation (engineering strain)	5.1	[73]
	Longissimus	N=7	$E_{single\ fiber} = 32.8 \pm 3.2$ kPa $E_{fiber\ bundle} = 62.9 \pm 14.7$ kPa	Linear elastic	Microindentation (engineering strain)	5.1	[73]
	Iliocostalis	N=7	$E_{single\ fiber} = 37.1 \pm 3.7$ kPa $E_{fiber\ bundle} = 58.8 \pm 7.7$ kPa	Linear elastic	Microindentation (engineering strain)	5.1	[73]
	BB, ECR, FCU, PT, SS, FDS, ECL *see legend at end of table*	N=21 (healthy) N=9 (spastic)	UTS = $150.4\ x10^3 \pm 25.7x10^3$ kPa (Healthy), $50.3\ x10^3 \pm 11.7x10^3$ kPa (Spastic)	Linear elastic	Tensile test (engineering strain)	5.1	[74]
	Medial Gastrocnemius muscle	N=6	G = 17.9-23 kPa (0°-30° plantarflexion)	Linear elastic	Ultrasound, shear-wave elastography (not reported)	5.1	[75]
	Deltoid (Regions A1, A2, M, P1, P2) ** *see legend at end of table*	N=8	**Elastic moduli (E, kPa) at 0 mm, 5 mm, 10 mm, and 15 mm of elongation** **A1:** 55.9 ± 8.9, 60.3 ± 11.1, 71.2 ± 10.8, 76.2 ± 11.6 **A2:** 72.4 ± 9.1, 77.6 ± 9.8, 109.9 ± 20.0, 129.3 ± 34.9 **M:** 63.0 ± 13.1, 69.7 ± 15.7, 97.4 ± 12.1, 123.5 ± 33.9 **P1:** 50.2 ± 9.9, 57.2 ± 17.5, 71.4 ± 14.9, 89.0 ± 27.1 **P2:** 39.1 ± 11.9, 42.8 ± 13.3, 52.2 ± 15.4, 61.3 ± 14.4	Linear elastic	Ultrasound, shear wave elastography (not reported)	5.1	[76]
Fat	Cadaver leg	N=70 N=13(m) N=11(f)	E = 23-61 kPa (m) E = 18-42 kPa (f)	Linear elastic	Compression test (not reported)	5.1	[71]
	Abdominal	N=9	**Biaxial (Triaxial) Test Parameter** C = 0.3 kPa (0.02 kPa) k_1 = 0.8 kPa (0.3 kPa) k_2 = 47.3 (40.0) κ = 0.09 (0.14) φ = 47.2° (51.4°)	Extended Neo-Hookean	Biaxial tensile test, triaxial shear test (stretch and right Cauchy-Green Deformation Tensor)	5.4	[77]
	Heel pad	N=10 (6 M, 4 F)	**350 mm/s** $C_{100} = C_{010} = 0.01\ x10^{-3}$ kPa $C_{110} = 0$ kPa $C_{200} = C_{020} = 0.1\ x10^{-3}$ kPa $\tau_1 = 0.5$s **175 mm/s** $g_1 = 28\ x10^{-3}$ kPa	Viscoelastic Mooney-Rivlin	Unconfined compression tests (stretch, deformation tensor not reported)	5.7	[78]

	Lower abdomen Subcutaneous Fat	N=19	E_{init} (kPa) = 1.6±0.8 E_{final} (kPa) = 11.7±6.4 A_1 (mN) = 1.7±0.5 A_2 (mN) = 4.2±2.8 A_3 (mN) = -0.4±3.2 k_1 (s^{-1}) = 0.004±0.003 k_2 (s^{-1}) = 0.0004±0.0001 t_1 (min) = 4.2±5.6 t_2 (min) = 41.7±16.7	Linear viscoelastic	Microscopy, tensile test (strain, strain definition not reported)	6.6	[79]
	Lower Abdomen Omental Fat	N=19	E_{init} (kPa) = 2.9±1.5 E_{final} (kPa) = 32±15.6 A_1 (mN) = 6.3±2.6 A_2 (mN) = 12.1±5.4 A_3 (mN) = 6.6±3.5 k_1 (s^{-1}) = 0.008±0.005 k_2 (s^{-1}) = 0.001±0.0003 t_1 (min) = 2.1±3.3 t_2 (min) =16.7±55.6	Linear viscoelastic	Microscopy, tensile test (strain, strain definition not reported)	6.6	[79]
Skin	Back	N=3 (1 M, 2 F)	UTS = $27.2x10^3$ ±$9.3x10^3$ kPa E = $98.97x10^3$ ±$97x10^3$ kPa	Linear elastic	Tensile test (stretch, deformation tensor not reported)	5.1	[24]
	N/A	N=8	C_1= 1.2-1.4 kPa C_2 = 960-$1.37x10^3$ kPa C_3 = 1.26-$14.5x10^{-4}$ kPa C_4 = 18.5–50.06	Anisotropic Veronda-Westmann	Tensile test (stretch, Right Cauchy-Green Deformation Tensor)	5.21	[80]
	Forearm, upper arm, shoulder, sternum, upper/lower back, abdomen, thigh, tibia, calf, knee	N=5	C_{1111} = 6.1-17.9 C_{1112} = .21-1.3 C_{1122} = 1.8-7.4 C_{1212} = 1.7-5.2 C_{1222} = .25-1.6 C_{2222} = 4.7-15.7 *Elasticity tensor entries were reported for each sample location.*	Linear elastic	2-D Tensile test (true strain)	5.1	[81]
	Forehead (fh), forearm (fa), Temporoparietal (tp), neck(n)	N=10	E = 330±40 kPa (fh) E = 650±50 kPa (tp) E = $1.03x10^3$±60 kPa (fa) E = $1.28x10^3$±60 kPa (n)	Linear elastic	Tensile test (strain, strain definition not reported)	5.1	[82]
	Thoracic, abdominal regions, dura mater, psoas major	N=10 (6 M, 4 F)	No material parameters reported	Linear viscoelastic	Tensile test (engineering strain)	N/A	[83]
	Back and abdomen	N=10	UTS = 2-$15x10^3$ kPa $E_{initial}$ = 100 kPa E_{final} = $18.8x10^3$ kPa $E_{viscuous}$ = $5.13x10^3$ kPa	Linear elastic	Strain rate test (engineering strain and true strain)	5.1	[84]
	Back	N=7 (3 M, 4 F)	**Parameters with angle relative to Langer Lines** **0°:** E = $95x10^3$ kPa, UTS = $23x10^3$ kPa **45°:** E = $90x10^3$ kPa, UTS = $23x10^3$ kPa **90°:** E = $50x10^3$ kPa, UTS = $15x10^3$ kPa **Parameters in different back locations** **Bottom:** E = $60x10^3$ kPa, UTS = $17x10^3$ kPa **Middle:** E = $90x10^3$ kPa, UTS = $25x10^3$ kPa **Top:** E = $80x10^3$ kPa, UTS = $24x10^3$ kPa	Linear elastic	Tensile test (stretch)	5.1	[85]

*Acronyms for ref. [60]: (BB)= biceps brachialis, (ECR)=extensor carpi radialis, (FCU)=flexor carpi ulnaris, (PR)= pronator teres, (SS)=subscapularis, (FDS)=flexor digitorum superficialis, (ECL)=extensor carpi radialis longus.

**Acronyms for ref. [62]: Regions of Deltoid muscle: anterior (A1, A2), middle(M), posterior (P1, P2)

3.1.1. Muscle

Ex vivo human muscle tissue were the focus of six studies [71–76]. Two investigated muscles from the leg, two from the arm and forearm, one from the shoulder, and one on muscle from the lower back. Three studies had less than ten samples [72,75,76], while the other three studies reported sample sizes greater than 20 [71,73,74], with the maximum number of samples being 70 [71].

3.1.2. Fat

Four studies were found that experimentally quantified the material parameters for human fat tissue [71,77–79]. Of these four studies, two tested fat from the abdomen, one the heel pad, and one tested on the leg. Sample sizes of ranged from nine to 19 samples.

3.1.3. Skin

Seven studies reported data from human skin samples [24,80–85]. These studies evaluated samples from the forehead, forearm, neck, back,

Table 2

In vivo human material properties. Each column provides information about the tissue type, sample location, sample size, testing position, parameters found, the model used, mechanical test (deformation definition in parentheses), where the foundational equation can be found in Tables 5 and 6, and the reference. *See acronyms used throughout the table after the last entry.

Tissue Type	Sample location	Sample size	Testing position	Parameters	Constitutive Law	Mechanical Test	Table 5/6	Ref.
Muscle	Medial gastrocnemius (MG), lateral gastrocnemius (LG), tibialis anterior (TA), soleus (S)	AB=4	Supine	**Parameters in different locations** **MG:** G = 24.86±0.71 kPa **LG:** G = 16.16±0.19 kPa **TA:** G = 12.03±0.39 kPa **S:** G = 16.77±0.24 kPa	Linear elastic	MRE, Shear wave elastography (not reported)	5.1	[33]
	Medial gastrocnemius (MG), lateral gastrocnemius (LG), tibialis anterior (TA), soleus (S)	DA=3	Supine	**Parameters in different locations** **MG:** G = 41.65±1.15 kPa **LG:** G = 39.28±0.90 kPa **TA:** G = 43.16±2.10 kPa **S:** G = 53.17±1.61 kPa	Linear elastic	MRE, Shear wave elastography (not reported)	5.1	[33]
	Brachialis: muscle fibers parallel (//) and perpendicular (⊥)	AB= 5 (M)	90° elbow flexion	**Elasticity ($\mu = \rho \cdot V_g^2$)** $\mu_{\parallel}$ = 5.86±0.20, 41.32±1.86, 57.58 ±0.91, 96.28±7.13, 100.80±3.22, $\mu_{\perp}$ = 1.58±0.15, 1.07±0.05, 1.97±0.52, 1.12±0.11, 1.12±0.21, **Viscosity** $\eta_{\parallel}$ = 0.65±0.10, 5.10±0.38, 6.72±0.19, 2.55±3.50, 3.80 ±2.15, $\eta_{\perp}$ = 0.92±0.06, 2.47±0.04, 2.89±0.33, 2.39±0.10, 2.50±0.29	Voigt	Ultrasound (strain, strain definition not reported)	6.11	[42]
	Quadriceps (Q), sartorius (S), gracilis (G)	AB=1 (M)	Supine	**Parameters in Different Locations** **Q:** C_{10} = 1.75 kPa, D = 18x10^{3} kPa^{-1} **G:** C_{10} = 2.2 kPa, D = 18x103 kPa^{-1} **S:** C_{10} = 3.75 kPa, D = 18x10^{3} kPa^{-1}	Neo-Hookean	Ultrasound, DIC (Deformation Gradient and Right Cauchy-Green Deformation Tensor)	5.2	[43]
	Biceps brachii (BB), gastrocnemius (G), flexor digitorum profundus (FDP), and soleus (S)	AB=12 (8 M, 4 F)	Seated	**Parameters in Different Locations** **BB:** 17.9±5.5 kPa **G:** 9.9±6.8 kPa **FDP:** 8.7±2.8 kPa **S:** 12.5±7.3 kPa	Linear elastic	MRE, Shear wave elastography (not reported)	5.1	[46]
	Buttocks	AB=1	Prone	μ_{1M} = 0.103x10^{-2}, μ_{2M} = .145x10^{-6} $G_{e,M}$ = 1.025 kPa $E_{e,M}$ = 3.1kPa D_{1M} = 0.195x10^{2}, D_{2M} = 0.166x10^{3} α_{1M} = 1.32, α_{2M} = -18.4	Viscoelastic Ogden	MRI (Left Cauchy-Green Deformation Tensor)	5.11	[47]
	Heel pad	AB= 6 (3M, 3 F)	Seated	**Mooney-Rivlin (5.6):** c_1 = -1.12x10^{-6} kPa, c_2 = 4.7x10^{-6} kPa **Generalized Rivlin (5.8):** c_{10} = 4.6x10^{-6} kPa, c_{20} = 2.34x10^{-7} kPa c_{30} = 2.3x10^{-6} kPa **Ogden (5.11):** μ_k = 10.2x10^{-6} kPa, α=8.04, c=10.2x10^{-6} kPa		Strain determined by geometric relationship (stretch, Hertz strain, volumetric strain)	5.6, 5.8, 5.11	[53]
	Medial gastrocnemius	AB=14 (11 M, 3 F)	Prone + supine	**Shear Modulus, Pre-Therapy (Post-Therapy)** **Therapy Group** $G_{neutral}$ = 19.0±5.9 (18.2±4.7) kPa $G_{dorsiflexion}$ = 112.3±39.5 (107.4±31.5) kPa **Control Group** $G_{neutral}$ = 19.2±6.0 (20.2±6.4) kPa $G_{dorsiflexion}$ = 108.4±40.0 (111.5±37.5) kPa	Linear elastic	Shear wave elastography, isokinetic dynamometer (not reported)	5.1	[86]
	Soleus	AB=14 (11 M, 3 F)	Prone + supine	**Shear Modulus, Pre-Therapy (Post-Therapy)** **Therapy Group** $G_{neutral}$ = 10.8±5.7 (11.0±4.3) kPa	Linear elastic	Shear wave elastography,	5.1	[86]

				G,,,.;n.,;,., - 24.8± I 1.8 (26.8±12.4) kPa Control Group G.,,mral - 8.6±2.6 (9.4±5.6) kPa Ga,,.;n,.:,,, - 22.1±9.2 (22.5±10.3) kPa		isokinetic dynamometer (not reported)		
	Posterior thigh	AB-5 (3 M,2 F)	Prone	Voigt (6.12): - 2.2-2.93x J0^{-3} kPa s, μ - 3.86-6.88 kPa Maxwell (6.15): - 18.6-55.1x I O^{+3} kPa s, μ - 4.62-7.32 kPa Zener (6.16): -3.96-6.65xJ0^{-3} kPa s, μ_1 -2.92-5.2 kPa, μ, - 3.34-7.38 kPa Springpot (6.17): - 3.96-6.65x I 0^{-3} kPa s, μ - 5.3-8.1 kPa, a - .158-.266 Rheo Spectris (MRE) Parameters G' - 1.835-1.853 (1.358-1.579) kPa G" - 0.1902-0.3154 (0.2129-0.4162) kPa		MRI, Shear wave elastography (not reported)	6.12, 6.15, 6.16, 6.17	[87]
	Quadriceps	AB-10 (M)	Seated	E,,,- 79-126 kPa	Voigt	Ultrasound, Shear wave elastography (true strain)	6.13	[88]
	Rectus femoris	AB-2(M)	Not reported	No material parameters reported	N/A	Sonoe\astography (not reported)	N/A	[89]
	Quadriceps (Q), sanorius (S), gracilis (G), lschio (l)	AB-I (M)	Supine	Parameters in different locations Overall: C10 = 11.6 k.Pa, D = 11.9xl0^{3} k.Pa^{-1} **Q:** C,o - 11.7 kPa, D - 16.3x 10^{3} kPa^{-1} **S/G:** C,o-21.3 kPa, D-12.9xl0^{3}kPa^{-1} I: C,o - 17.3 kPa, D - 12.9x I 0^{3} kPa^{-1}	Neo-Hookean	Ultrasound, DIC (Deformation Gradient)	5.2	[90]
	Gluteal Tissue	AB-I (M)	Prone	Goo,M- 0.881 kPa; Go,M - 1.032 kPa Koo,M - 77 kPa Time-Independent Constants c,M - 0.441 kPa D,M-2.59xl0^{-2} M·lkPa^{-1} k_{1M} = 13.1 kPa k,M-4,69xl0^{1} KM- J/3 Time-Dependent Constants g, - 7.67x I 0^{-1}, k, - 1.17x10^{-2}, r, - 2 s g, - 6.44x 10^{-2}, k_2- 2.29x IO^{-1}, r, - 40 s g_3-6.08xl0^{-1}, k_3-3.74xl0^{4}, r_3-80 s g,- 2.18x10^{-2}, I<, - 1.29xJ0^{-5}, r, - 200 s	Viscoelastic Holzapfel-Gasser-Ogden,	MRI (modified right Cauchy-Green Deformation Tensor)	5.15	[91]
	Biceps brachii	AB-I	Lifting 6 kg charge	E - 8.29±0.14 kPa Vs-^{1} =12 ms^{-1}, V/ =3 ms^{-1}	Linear elastic	Ultrasound, Shear wave elastography (not reported)	5.1, 6.21	[92]
	Biceps brachii	N-1	N/A	E, - I3±4.5 kPa E, - I85±60 kPa Shear modulus (μ13)- 54±3 kPa Experimental shear modulus (μ12) = 3.4±1 I kPa	Transversely isotropic linear elastic	MRE, Shear wave elastography (Green-Lagrange Strain Tensor)	5.1	[93]
	Biceps brachii	AB-5 (M)	Supine with elbow angle at 90°	muscle stiffness= 14.5±1.77 kPa/kg	Linear elastic	MRE, Shear wave elastography (not reported)	5.1, 6.1	[94]
Fat	Quadriceps (Q), sartorius (S), gracilis (G)	AB-I (M)	N/A	Undifferentiated fat tissues of thigh C,o - 0.64 kPa D - 29.4x I 0^{3} kPa^{-1} Isolated fat tissue of muscle regions: C,o-0.52 kPa D - 30.5x I 0^{3} kPa^{-1}	Neo-Hookean	Ultrasound, DIC (Deformation Gradient)	5.2	[90]
Skin Hypodermis=HD, Dermis -D	Inner forearm	N-I0(M)	Seated	E-4.5-8 kPa ** depending on various factors, including indentation depth and assumed skin thickness*	2-layer linear elastic model	Indentation (displacement)	5.23-5.25	[39]
	Forearm	AB-10	Seated	Cto - 9.4±3.6 kPa C_{1-1} 82±60 kPa	Generalized Mooney	Ultrasound, Suction (displacement)	5.5	[40]
	Shoulder, Thigh, Calf	AB-30 (F)	N/A	No material parameters reported	N/A	Ultrasound and laser	N/A	[41]

						displacement for thickness and strain (not reported)		
	Forearm Skin	AB 20 (F)	*NIA*	E• 8.3±2.1 kPa G• 2.8±0.46 kPa	Linear elastic	Indentation (displacement)	5.22	[95]
	Forearm	AB J (M)	*NIA*	**Hypodermis:** C10 0.06 kPa, K 0 kPa Epidermis+ Dermis: Cio = 0.80 kPa, K = 0 kPa	Neo-Hookean	MRJ (Deformation Gradient)	5.3	[96]
	Volar forearm	AB 3 (F)	Seated	Coefficient of friction = 0.7	Friction	Indentation (displacement)	6.18	[97]
	Anterior forearm	AB J0 (M)	NIA	E 129±88 kPa	Linear elastic	Ultrasound (strain, strain definition not reported)	5.1	[98]
	Palmar skin, dorsal forearm skin, ventral forearm skin	AB 17	*NIA*	Rate of energy damping (*a*, time dependent) **Palm:** *a* 0.077±0.058 Dorsal Forearm: a= 0.049±0.008 Ventral Forearm: a= 0.044±0.006	*NIA*	Ballistometer (displacement)	*NIA*	[99]
	Forearm	AB 138	NIA	Younger (<30 years): E 420 kPa **Older (>30 years):** E 850 kPa	Linear elastic	Torque measurement (strain, strain definition not reported)	6.1	[100]
	PH, DH,AF, PF, AL, PL ** *see legend at end of table*	AB J0	*NIA*	Static coefficient of friction (μ) **PH:** μ 0.62±0.22 **OH:μ** 0.47±0.12 **AF:** μ 0.46±0.10 **PF:** μ 0.43±0.10 **AL:** μ 0.40±0.10 **PL:** μ 0.40±0.09	Friction	Probe indentation (displacement)	6.19	[IOI]
	Arm and leg	AB 95 (47 **M,** 48 F)	Supine	Only plots and soft tissue thickness reported	Linear elastic	Ultrasound, Indentation (displacement)	5.1	[102]
	Forearm, dorsal forearm, calf, thigh, back, palm of hand	N JJ (II M)	*NIA*	Parameters in different locations **Forearm:** μ 11.13±5.62 kPa, 8.96x I Q^{-3}±6.06x I Q^{-3} kPa's **Dorsal Forearm:μ** 14.50±9.34 kPa, 10.09xJ0^{-3}±10.37xl0^{-3} kPa*s **Calf:μ** 22.33±8.77 kPa, 7.76xl0^{-3}±3.77xlQ^{-3} kPa's **Thigh:** 11 11.37±5.41 kPa, I 1.04xl0^{-3}±12.97xl0^{-3} kPa*s **Back:** μ 6.63±3.40 kPa, 8.16x I 0^{-3}±7.31x I 0^{-3} kPa's **Palm:μ** 28.41±13.32 kPa, 32.94xlQ^{-3}±22.97xl0^{-3} kPa's	Voigt Model	Scanning laser vibrometer (displacement)	6.11	[103]
Bulk Tissue	Buttocks and Thighs [Muscle +Skin+ Fat]	AB 20 (10 M, 10 F)	Seated, Quadruped, and Prone	Parameters in different positions **Seated:μ** 4.8-6 kPa, *a* 3.5-7.5 **Quadruped:** μ 4.6-6.5 kPa, *a* 3-6 **Prone:** μ 2.5-4 kPa, *a* 9.5-13.5	Ogden	Indentation (stretch)	5.10	[23]
	Buttocks and Thighs [Muscle +Skin+ Fat]	AB 20 (10 M, 10 F), DA IJ (9 M,2 F)	Quadruped	Parameters in different populations **DA Males:μ** 3.51-4.54 kPa, *a* 4.42--6.37 **DA Females:** μ 2.31-3.95 kPa, *a* 4.98-6.40 **AB Males:** μ 4.78-5.35 kPa, *a* 3.84-5.72 **AB Females:** μ 4.39-6.65 kPa, *a* 3.26-5.53	Ogden	Indentation (stretch)	5.10	[34]
	Buttocks [Skin+Fat]	AB J	Prone	μ, 0.1 I 8x I Q^{-2}; μ, 0.644x I Q^{-7} *a*, -0.1076; *a*, -0.03189x10^{2} D_1 0.169x 10^{2}; D_2 0.477x 10^{1} *G* 1.182 kPa; E 3.53 kPa	Ogden	MRI (Left Cauchy-Green Deformation Tensor)	5.11	[47]
	Lower back, buttocks, thighs [Muscle +Skin+ Fat]	AB 20 (10 M, 10 F)	Seated	Parameters in different locations, male (female) **LB:** c, 0.2 kPa (0.04 kPa), c, 9.9 kPa (4.9 kPa) **PB:** c, 2. Ix JQ^{-6} kPa (0.39 kPa), c_2 11.3 kPa (7.2 kPa) **18:** c, 0.81 kPa (1.4 kPa), c, 2.9 kPa (2.7 kPa) **PT:** c, 3.2x JQ^{-5} kPa (l.2x I Q^{-5} kPa), c, 22.8 kPa (18.6 kPa) **MT:** c, 1.6 kPa (1.0 kPa), c, 5.3 kPa (4.5 kPa) **OT:** c, 3.4 kPa (2.2 kPa), c, 0.73 kPa (1.3 kPa) ****see legend at end of table*	Mooney-Rivlin	Motion Capture (Right Cauchy-Green Deformation Tensor)	5.6	[48]

	Thigh [Muscle +Skin + Fat]	AB=20 (10 M, 10 F)	Seated	**Semi-3D:** C_1 = 3.6-11.9 kPa, K = 0 kPa **3D:** C_1 = 3.02-8.5 kPa, K = 0 kPa	Neo-Hookean	Motion Capture (Right Cauchy-Green Deformation Tensor)	5.3	[49]
	Thigh [Muscle +Skin + Fat]	AB=20 (10 M, 10 F)	Seated	**Semi-3D:** μ = 2.3–3.4 kPa, α = 3.5-13.0, K = 0 kPa **3D:** μ = 0.49-1.7 kPa, α = 5.4-35.8, K = 0 kPa	Ogden	Motion Capture (Deformation Gradient)	5.12	[49]
	Thigh [Muscle +Skin + Fat]	AB=20 (10 M, 10 F)	Seated	**Semi-3D:** C_1 = 1.7-2.2 kPa, C_2 = 1.8-9.7, K = 0 kPa **3D:** C_1 = 1.5-2.2 kPa, C_2 = 0.62-6.0, K = 0 kPa	Mooney-Rivlin	Motion Capture (Deformation Gradient, Right Cauchy-Green Deformation Tensor)	5.6	[49]
	Thigh [Muscle +Skin + Fat]	AB=20 (10 M, 10 F)	Seated	**Semi-3D:** E = 3.2-18.3 kPa, c = 0.95-16.3 kPa **3D:** E = 2.7-25.3 kPa, c = 3.2-23.3 kPa	Fung	Motion Capture (Deformation Gradient, Right Cauchy-Green Deformation Tensor)	5.9	[49]
	Skin+fat	N=1	Prone	No material parameters reported	N/A	MRI, indentation (not reported)	N/A	[50]
	Arm and leg [Muscle +Skin + Fat]	AB=100 (50 M, 50 F)	Supine	No material parameters reported	N/A	Ultrasound, Indentation (not reported)	N/A	[51]
	Gluteal Tissue [Skin+Fat]	AB=1 (M)	prone	$G_{\infty,S/F}$ = 1.854 kPa; $G_{0,S/F}$ = 1.92 kPa $K_{\infty,S/F}$ = 719 kPa **Time-independent Constants** $c_{1S/F}(kPa) = 0.928$ $D_{S/F}(M^{-1}kPa^{-1}) = 2.78x10^{-3}$ $k_{1S/F}\ (kPa) = 6.56x10^{-6}$ $k_{2S/F} = 1.51x10^{2}$ $\kappa_{S/F} = \frac{1}{3}$ **Time-Dependent Constants** g_1 = 1.33x10^{-2}, k_1 = 1.32x10^{-2}, τ_1 = 2 s g_2 = 3.64x10^{-3}, k_2 = 3.25x10^{-4}, τ_2 = 40 s g_3 = 3.85x10^{-4}, k_3 = 5.60x10^{-4}, τ_3 = 80 s g_4 = 1.60x10^{-2}, k_4 = 1.89x10^{-5}, τ_4 = 200 s	Viscoelastic Holzapfel-Gasser-Ogden,	MRI (modified right Cauchy-Green Deformation Tensor)	5.14	[91]

* AB = able-bodied, DA = disabled, MRE = magnetic resonance elastography, MRI = magnetic resonance imaging, DIC = digital image correlation
** Acronyms for ref. [78]: PH = palm of hand, DH = dorsum of the hand, AF = anterior side of the forearm, PF = posterior side of the forearm, AL = anterior leg, PL = posterior leg, E*= reduced young's modulus, E_f*=Young's modulus of the skin
***Acronyms for ref. [35]: LB = lower back, PB = posterior buttocks, IB = inferior buttocks, PT = proximal thigh, MT = middle thigh, DT = distal thigh

and abdomen. Three studies reported a sample size of at least 10 [82–84], one had eight samples [80], one had seven samples [85], one tested five samples [81], and one tested three samples [24].

3.2. *In vivo human tissue properties*

34 studies were found that included *in vivo* properties of human muscle, fat, skin, and bulk tissue (defined as a homogeneous material of muscle, fat, and skin by lumping the parameters), as presented in Table 2. Factors that differed between the studies included region of the body tested, number of participants, body position, and whether the study included able-bodied people or persons with disabilities.

3.2.1. *Muscle*

Material property values were obtained from multiple body regions in multiple body positions and from both able-bodied people and those with disabilities. 12 studies reported tissue parameters from muscles of the lower limb, including the gastrocnemius, gracilis, sartorius, quadriceps, posterior thigh, heel pad of the foot, tibialis anterior, rectus femoris, buttocks, and triceps surae [33,43,44,46,47,53,86–91]. Five studies reported on tissue from the upper limb including the biceps brachii, flexor digitorum profundus, and brachialis [42,46,92–94]. Six studies were conducted while participants were in the supine position (lying on their back) [33,43,44,86,90,94], three studies were conducted in the seated position [46,53,88], and four studies the prone position (lying on their stomach) [47,86,87,91]. Four studies did not report the testing position or tested in a different manner than supine, prone or seated [42,89,92,93]. Three studies included able-bodied sample sizes of 10 or greater [46,86,88], and 11 studies tested able-bodied participants with a sample size less than 10 [33,42,43,47,53,87,89–94]. One study included individuals with physical disabilities with a sample size less than 10 [33], and two studies did not report whether the

Table 3

Ex vivo Animal Material Properties *acronyms used in this Table are written out after the last entry. Each column provides information about the tissue type, animal, sample location, sample size, parameters found, the model used, mechanical test (deformation definition in parentheses), where the foundational equation can be found in Tables 5 and 6, and the reference. *See acronyms used throughout the table after the last entry.

Tissue Type	Animal	Sample Location	Sample Size	Parameters	Constitutive Law	Mechanical Test	Table 5/6	Ref.
Muscle	Pig	Buttock	N=17 N=19 N=17 N=23 N=24 N=23	**Non-Preconditioned Parameter (Preconditioned Parameter)** **G_∞ (kPa)** = 0.58±.17 (0.81±0.41) **G_S (kPa)** = 8.51±3.9 (5.71±1.79) **G_L (kPa)** = 0.61±0.22 (0.81±0.41)	Linear viscoelastic	Indentation (displacement)	6.5	[22]
	Frog	N/A	N=7	**Resting Parameter (Contracted Parameter)** **E (kPa)** = $2.64x10^6$ ($2.65x10^6$) **E_t (kPa)** = $4.40x10^6$ ($4.57x10^6$) **G (kPa)** = $3.39x10^6$ ($3.43x10^6$)	Linear anisotropic	Ultrasound (not reported)	5.1	[45]
	Pig	Spine region and abdomen	N=1	**Mooney-Rivlin (5.6):** C_1 = 190 kPa, C_2 = -210 kPa, K = 0 **Yeoh (5.15):** C_1 = 4.2 kPa, C_2 = 51 kPa, C_3 = -13 kPa **Neo-Hookean (5.3):** C_1 = 54 kPa, K = 0 **Ogden (5.12):** μ = 5 kPa, α = 5.7, K = 0 **Humphrey (5.16):** C_1 = 9 kPa, C_2 = 1.5 **Martins (5.18):** C_1 = 540 kPa, C_2 = -0.22, C_3 = 790 kPa, C_4 = 0.43 **Veronda-Westmann (5.19):** C_1 = 21 kPa, C_2 = 1.1		Uniaxial tensile test (Right Cauchy-Green Deformation Tensor and Left Cauchy-Green Deformation Tensor)	5.6, 5.16, 5.3, 5.12, 5.17, 5.19, 5.20	[55]
	Cow	Beef sirloin	N=2	No parameters reported	N/A	Sonoelastography (not reported)	N/A	[89]
	Cow	Peroneus tertius and extensor digitorum longus	N=10	G = 23.8±6.68 kPa	Linear elastic	MRE, Shear wave elastography (not reported)	5.1	[94]
	Pig	Semitendinosus	N=5	No (muscle) parameters reported	Linear elastic	MRE, Shear wave elastography (not reported)	5.1	[104]
	Pig	Pelvic limb	N=4	c_1 = 0.948±.024 c_2 = 11.36±0.085 c_3 = .880±.009 c_4 = -.070±0.176 c_5 = 2.873±0.039	Humphrey	Uniaxial unconfined compression (stretch, true strain)	5.18	[105]
	Pig	Buttock	N=3	Stiffness (S) = 8.5-347 kN/m Damping (D) = 0-556 Ns/m	Kelvin–Voigt	Cyclic compression (not reported)	6.14	[106]
	Rabbit	Extensor digitorum longus	N=6	$E_{longitudinal}$ = 447±97.7 kPa $E_{transverse}$ = 24.4±14.7 kPa G = 3.87±3.39 kPa	Linear elastic	Tension test (Green Strain)	5.1	[107]
	Cow	Rectal wall	Transverse, N=2; Longitudinal, N=1	E = 14 kPa v = 0.495 (for all)	Linear elastic	Compression test, analog displacement measurements (strain, strain definition not reported)	5.1	[108]
Fat	Pig	Buttock	N=3	G_d = 14.9±4.8 kPa	Linear viscoelastic	Oscillatory shear experiment, frequency sweep (strain, strain definition not reported)	6.7	[109]
	Sheep	Buttock	N=20	**Non-Preconditioned Parameter (Preconditioned Parameter)** H_{Long} (Pa) = 10.6±4.2 (10.1±4.0) H_{short} (Pa) = 28.9±14.9 (18.2±6.9) E^{indent} (Pa) = 860±440 (840±200) E^{swift} (Pa) = 22.6±10 (16.3±9.9)	Linear Viscoelastic	Distance measurements (strain, strain definition not reported)	6.2, 6.5	[110]
Skin	Mouse	Back	N=14	C_1 = 0.278±0.118 kPa C_2 = 10.2±2.7 C_3 = 4.32-5.28 kPa C_4 = 21.1-29.0 kPa *C5 and C6 are dependent on the other constants	Anisotropic Veronda-Westmann	Tensile test (stretch, Right Cauchy-Green Deformation Tensor)	5.21	[80]

	Pig	Belly	N=1	**Load rate dependency (0.25-10 % s^{-1})** μ = 0.2-3.1 kPa α = 7.772-10.46 **Temperature dependency (10-60°)** μ = 0.368-3.909 kPa α = 8.504-9.677	Ogden	Tensile test (stretch)	5.13	[111]
	Cow	Buttock	N=4	μ = 400 kPa α = 4.6 K = 0	Ogden	Uniaxial tensile test (stretch)	5.12	[112]
	Pig	Abdomen, Spine	N=1	**Abdomen** **Neo-Hookean:** C_1 = 0.28 **Mooney-Rivlin:** C_1 = 1.06, C_2 = -1.57 **Ogden:** C_1 = 0.06, C_2 = 7.73 **Yeoh:** C_1 = -0.02, C_2 = 0.05, C_3 = 0.003 **Humphrey:** C_1 = 0.13, C_2 = 0.55 **Martins:** C_1 = 0.87, C_2 = 0.45, C_3 = 1.41, C_4 = -0.62 **Veronda-Westmann:** C_1 = 0.47, C_2 = 0.40 **Spine** **Neo-Hookean:** C_1 = 5.40 **Mooney-Rivlin:** C_1 = 80.5, C_2 = -82.3 **Ogden:** C_1 = 3.58, C_2 = 27.4 **Yeoh:** C_1 = 0.69, C_2 = 104, C_3 = -400 **Humphrey:** C_1 = 0.09, C_2 = 26.8 **Martins:** C_1 = 3.13, C_2 = -10.5, C_3 = 12.5, C_4 = 7.98 **Veronda-Westmann:** C_1 = 0.35, C_2 = 15.0 ** for units and material model equivalents of model constants, see after the last entry in the table		Uniaxial tensile test (engineering strain, stretch, Cauchy-Green Deformation Tensor – orientation not reported)	5.3, 5.6, 5.11, 5.16, 5.17, 5.19, 5.20	[113]
	Pig	Abdomen	N=1	C_0 = 50 m/s S = 4.3	Hugonoit relationship	Uniaxial tensile testing, DIC (true strain)	6.26	[114]
	Rat	Back	N=63 (M)	**1 month** Tangent Modulus = $14x10^3$ kPa **4 months** Tangent Modulus = $36x10^3$ kPa	Elastic with a statistical recruitment of fibers	Tensile test (displacement, engineering strain)	6.4	[115]
	Rat	Posterior hind limbs	N=30 (M)	Stiffness = 2.3-4.4 N/mm	Linear elastic	Tensiometer (displacement)	5.1	[116]
	Pig	Buttock	N/A	**Parameters at different strain rates** **0.004 s^{-1}:** μ (kPa) = 400, α = 12, K = 0 **0.4 s^{-1}:** μ (kPa) = $1.2x10^3$, α = 12, K = 0 **40 s^{-1}:** μ (kPa) = $2.2x10^3$, α = 12, K = 0 **4000 s^{-1}:** μ (kPa) = $7.5x10^3$, α = 12, K = 0	Ogden	Uniaxial compression (engineering strain, stretch)	5.12	[117]

* MRE= magnetic resonance elastography, G_s= shear modulus, G_1=long-term shear modulus, G(t)=transient shear modulus, E^{indent}=short term elastic moduli, G_d=dynamic shear modulus

** C_1 is C_{10} for the Neo-Hookean model in MPa; C_1 is C_1 for the Mooney-Rivlin model in MPa, C_2 is C_2 for the Mooney-Rivlin model in MPa; C_1 is μ for the Ogden model in MPa, C_2 is α for the Ogden model; C_1 is C_1 for the Yeoh model in MPa, C_2 is C_2 for the Yeoh model in MPa, C_3 is C_3 for the Yeoh model in MPa; C_1 is C_1 for the Humphrey model in MPa, C_2 is C_2 for the Humphrey model; C_1 is C_1 for the Martins model in MPa, C_2 is C_2 for the Martins model, C_3 is C_3 for the Martins model in MPa, C_4 is C_4 for the Martins model; C_1 is C_1 for the Veronda-Westmann model in MPa, C_2 is C_2 for the Veronda-Westmann model

participants were able-bodied or had a disability [92,93].

3.2.2. Fat

Fat tissue was included in one study that only included able-bodied participants [90]. The sample location was not identified, and one male was tested in the supine position.

3.2.3. Skin

In vivo samples of human skin from several regions of the body have been tested. The testing regions included the forearm, palm, back, shoulder, thigh, and calf, leading to a range of tissue parameters [39–41,95–103]. Testing positions varied as well, as three studies tested while seated [39,40,97], and eight studies did not report the testing position [41,95,96,98–101,103]. None tested individuals with disabilities. Nine studies included at least 10 able-bodied individuals, with some testing up to 138 participants [39–41,95,98–103]. Two studies tested less than five participants [96,97].

3.2.4. Bulk tissue

While some *in vivo* experiments separated the muscle, fat, and skin into separate components, there were several that described the tissues as a homogeneous material, using lumped parameters. All studies that described the bulk soft tissue behavior in humans collected data from the buttocks and thigh regions. Five of these studies had at least 20 participants [23,34,48,49,51], while three had a single participant [47,50,91]. Three studies determined tissue properties in the seated position [23,48,49], four in the prone position [23,47,50,91], two in the quadruped position [23,34], and one in the supine position [51]. Eight studies included data from able-bodied people, while one study reported material parameters in people with spinal cord injuries and able-bodied people [23,34,47–51,91,102].

3.3. Using animal properties for human modeling data

26 studies on *in vivo* and *ex vivo* animal muscle, fat, and skin tissue were published between 1987 and 2014, but the authors found no more

Table 4

In vivo Animal Material Properties. Each column provides information about the tissue type, animal, sample location, sample size, test position, parameters found, the model used, mechanical test (deformation definition in parentheses), where the foundational equation can be found in Tables 5 and 6, and the reference. *See acronyms used throughout the table after the last entry.

Tissue Type	Animal	Sample location	Sample size	Test position	Parameter	Constitutive Law	Mechanical Test	Table 5/6	Ref.
Muscle	Pig	Buttock	N=1	Prone	μ = 6 kPa α = 5 K = 0 G = 15 kPa	Ogden	MRI (stretch, Green-Lagrange Strain Tensor)	5.12	[5]
	Rat	Hind limb muscles	N=4	Supine	No material parameters reported	Neo-Hookean	MRI, indentation (Left Cauchy-Green Deformation Tensor)	N/A	[18]
	Rat	Gracilis, tibialis anterior	N=33	Prone	No material parameters reported	Polynomial Stress Relationship	Compression, uniaxial tensile test (stretch, true strain)	6.20	[19]
	Rat	Tibialis anterior	N=2	N/A	μ = 15.6±5.4 kPa α = 21.4 ± 5.7 δ = 0.549±0.056 τ = 6.01±0.42 s G = 15.6±5.4 kPa	Viscoelastic Ogden	Indentation, compression (stretch, Green-Lagrange Strain Tensor)	5.14	[20]
	Rat	Gracilis	N=32	Prone	G_∞ = 0.348-0.730 kPa	Linear Viscoelastic	Constant pressure, displacement transducer (displacement)	6.8	[118]
	Rat	Tibialis anterior	N=15	N/A	No material parameters reported	N/A	Waveform, motion analysis (engineering strain)	N/A	[119]
	Rabbit	Tibialis anterior and extensor digitorium longus	N=7	N/A	C = 0.62±0.08 kPa τ_1 = 0.012±0.005 τ_2 = 300 s	Quasi-Linear Viscoelastic	Force-displacement (displacement, strain, strain definition not reported)	6.9	[120]
Fat	Pig	Buttock	N=1	Prone	μ = 3 kPa α = 5 K = 0 G = 7.5 kPa	Ogden	MRI (stretch, Green-Lagrange Strain Tensor)	5.12	[5]
	Rat	Gracilis (Gr), tibialis anterior (TA)	N=33	Prone	No material parameters reported	Polynomial relationship	Compression in vivo, uniaxial tensile test (stretch, true strain)	6.20	[19]
Skin	Pig	Buttock	N=1	Prone	μ = 8 kPa α = 5 K = 0 G = 20 kPa	Ogden	MRI (stretch, Green-Lagrange Strain Tensor)	5.12	[5]
	Rat	Gracilis (Gr), tibialis anterior (TA)	N=33	Prone	No material parameters reported	Polynomial relationship	Compression, uniaxial tensile test (stretch, true strain)	6.20	[19]
	Pig	Abdomen, shoulder	N=2	N/A	No material parameters reported	N/A	Extensometer - compression measurements (displacement)	N/A	[21]
	Rat	Posterior hind limbs	N=10 (M)	N/A	Stiffness = 14.8-52.1 N/m	Linear elastic	Optical coherance tomography-based air-jet indentation (displacement)	5.1	[116]

* MRI= magnetic resonance imaging

recent studies. Samples were taken from sheep, pigs, mice, rats, rabbits, cows, and frogs.

3.3.1. Ex vivo

The studies on *ex vivo* animal tissue sample included properties from muscle, fat, and skin. The animals from which the tissue samples were taken are identified in Table 3. Muscle tissue data were collected from the extensor digitorum longus, peroneus tertius, thoracic muscle, gluteus, semitendinosus, abdomen, and pelvic muscles [22,45,55,89,94,104–108]. Two studies included sample sizes of 10 or more [22,94]. Eight studies included a sample size less than 10 [45,55,89,104–108]. Two studies included fat properties taken from the gluteus, with sample

Table 5
Strain energy density equations to determine material parameters.

No.	Constitutive Law	Strain Energy Density Equation
1	Linear Elastic (Hookean)	$W(\epsilon_{ij}) = \frac{1}{2}\sum_{i=1}^{3}\sum_{j=1}^{3}\sigma_{ij}\epsilon_{ij}$
2	Neo-Hookean (1)	$W(I) = C_{10}(\bar{I}_1 - 3) + \frac{1}{D^{NH}}(J-1)^2$
3	Neo-Hookean (2)	$W(I) = C_{10}(I_1 - 3) + \frac{1}{2}K^{NH}(lnJ)^2, K^{NH} = \frac{2C_{10}}{1-2v}$
4	Extended Neo-Hookean	$W(I) = \frac{C}{2}(I_1 - 3) + \frac{k_1}{k_2}(e^{k_2[\kappa I_1 + (1-3\kappa)I_4 - 1]^2} - 1)$
5	Generalized Mooney	$W(I) = C_{10}(I_1 - 3) + C_{11}(I_1 - 3)(I_2 - 3)$
6	Mooney-Rivlin	$W(I) = C_1(I_1 - 3) + C_2(I_2 - 3) + \frac{1}{2}K^{MR}(lnJ)^2, K^{MR} = \frac{2(C_1+C_2)}{1-2v}$
7	Viscoelastic Mooney-Rivlin	$W(I_1, I_2) = \sum_{i+j=1}^{N} C_{ij0}(I_1 - 3)^i(I_2 - 3)^j\left(1 - \sum_{k=1}^{n} g_k\left(1 - e^{-\frac{t}{\tau_k}}\right)\right)$
8	Generalized Rivlin	$W(I) = \sum_{i+j=1}^{3} c_{ij}(I_1 - 3)^i(I_2 - 3)^j$
9	Fung	$W(I,E) = \frac{1}{2}C^F(e^Q - 1) + \frac{1}{2}K^F(lnJ)^2, K^F = \frac{E^F}{3(1-2v)}$ $Q = \gamma^F(I:E^F)^2 + \frac{\gamma^F(1-2v)}{v}I:E^2$ $\gamma^F = \frac{E^F v}{C^F(1+v)(1-2v)}$
10	Ogden (1)	$W(\lambda) = \sum_{i=1}^{N}\frac{\mu_i}{\alpha_i}(\lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3) + \sum_{i=1}^{N}\frac{1}{D_i}(J-1)^2$
11	Ogden (2)	$W(\lambda) = \sum_{k=1}^{N} 2\frac{\mu_k}{\alpha_k^2}(\lambda_1^{-\alpha_k} + \lambda_2^{-\alpha_k} + \lambda_3^{-\alpha_k} - 3) + \sum_{k=1}^{N}\frac{1}{D_k}(J-1)^{2k}$
12	Ogden (3)	$W(\lambda) = \sum_{i=1}^{N}\frac{\mu_i}{\alpha_i}(\lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3) + \frac{1}{2}K(lnJ)^2$
13	Ogden (4)	$W(\lambda) = \sum_{k=1}^{N} 2\frac{\mu_k}{\alpha_k^2}(\lambda_1^{-\alpha_k-1} + \lambda_2^{-\alpha_k-1} + \lambda_3^{-\alpha_k-1} - 3) + \sum_{k=1}^{N}\frac{1}{D_k}(J-1)^2$
14	Viscoelastic Ogden	$W(\lambda) = \frac{\mu}{\alpha}(\lambda_1^{\alpha} + \lambda_2^{\alpha} + \lambda_3^{\alpha} - 3)$

		$S(\zeta,t) = (1-\delta)\frac{\partial W}{\partial E} + \int_0^t \delta \frac{\partial W}{\partial E} e^{-\frac{t-\zeta}{\tau}} d\zeta$
15	**Viscoelastic Holzapfel-Gasser-Ogden**	$W(C,H_i) = c_1(\bar{C_1}-3) + \frac{k_1}{2k_2}\sum_{i=1}^{N}(e^{k_2\bar{E}_i^2}-1) + \frac{1}{D}\left[\frac{1}{2}(J^2-1) - \ln J\right]$ $= G_0\left[1 - \sum_{i=1}^{N_G} g_i\left(1 - e^{-(t/\tau_i^G)}\right)\right]$ $K(t) = K_0\left[1 - \sum_{i=1}^{N_K} k_i\left(1 - e^{-(t/\tau_i^K)}\right)\right]$
16	**Yeoh**	$W(I_1) = C_1(I_1-3) + C_2(I_1-3)^2 + C_3(I_1-3)^3$
17	**Humphrey (1)**	$W(I_1) = C_1(e^{C_2(I_1-3)}-1)$
18	**Humphrey (2)**	$W(I_1,\lambda) = c_1(\lambda-1)^2 + c_2(\lambda-1)^3 + c_3(I_1-3) + c_4(I_1-3)(\lambda-1) + c_5(I_1-3)^2$
19	**Martins**	$W(I_1,\lambda) = C_1\left(e^{C_2(I_1-3)}-1\right) + C_3(e^{C_4(\lambda-)^c}-1)$
20	**Veronda-Westmann**	$W(I_1,I_2) = C_1\left(e^{C_2(I_1-3)}-1\right) - \frac{C_1C_2}{2}(I_2-3)$
21	**Anisotropic Veronda-Westmann**	$W(I) = C_1\left[e^{C_2(I_1-3)}-1\right] - \frac{C_1C_2}{2}(I_2-3)\ for\ \lambda < 1$ $W(I) = C_1\left[e^{C_2(I_1-3)}-1\right] - \frac{C_1C_2}{2}(I_2-3) + C_3\left(e^{C_4(\lambda-1)}-1\right)\ for\ 1 < \lambda < \lambda_m$ $W(I) = C_1\left[e^{C_2(I_1-3)}-1\right] - \frac{C_1C_2}{2}(I_2-3) + C_5 + C_6\lambda\ for\ \lambda \geq \lambda_m$ *C5 and C6 are dependent on the other constants

sizes of three and 20 [109,110]. Eight studies reported properties from skin samples, where four studies had sample sizes less than five [111–114], three studies had 14 or more samples [80,115,116], and one study did not report a sample size [117]. Skin samples were taken from the abdomen, posterior hindlimb, back, and gluteus.

3.3.2. *In vivo*

In vivo studies found on animals include seven that tested muscle [5, 18–20,118–120] and two that tested skin [21,116], with two reporting on fat properties [5,19], which are listed in Table 4. Studies of animal muscle tissue included the tibialis anterior, gluteus, extensor digitorum longus and gracilis. In addition to testing occurring over multiple body regions, body position of the animals varied. Three investigations tested in the prone position [5,19,118], one in the supine position [18], and five did not report the testing position [20,21,116,119,120]. Among the studies, three had a sample size greater than 10 [19,118,119], and six studies had a sample size of 10 or less [5,18,20,21,116,120].

3.4. *Material model choices*

The choice of material model used to represent human tissue determined the reported material parameters. Human tissues were described using multiple different material models to represent the mechanical properties of muscle, fat, skin, and bulk tissue. 11 studies used a linear viscoelastic model with time-dependent behavior for muscle, fat, skin and bulk tissue [22,79,83,84,88,99,106,109,110,118,120]. Six studies used hyperelastic and viscoelastic terms to generate material models [20,47,77,78,87,91]. Additionally, 16 studies used a purely hyperelastic modeling approach for muscle [5,18,19,43,53], skin [40,96,111–113, 117], or bulk tissues [23,34,47–49]. Four studies investigated the differences between applying multiple material models in describing soft tissue, demonstrating the various material parameters that may be obtained for the experimental data [49,53,55,113]. Several other studies reported using linear elastic models to represent skin [24,39,81, 82,84,95,98,100,102,115,116], fat [71], and muscle [33,45,46,71, 73–76,86,92–94,104,107,108].

3.5. *Measures of deformation in tissue*

Raw displacement, strain, or stretch data in the tissues were reported. Displacement was the *absolute magnitude* of the change in tissue thickness when loaded versus unloaded. Tissue strain was the *ratio of the change* in thickness of tissue while loaded relative to its thickness in a non-loaded condition, while tissue stretch was the *ratio of the tissue thickness* in a loaded condition relative to a non-loaded one. Several definitions for deformation were used to determine tissue properties (i.e. Left Cauchy-Green Deformation Tensor, engineering strain, stretch, displacement, etc.). The mechanical test used to measure the deformation used by the authors is reported in Tables 1–4 in the 'Mechanical Test' column.

Measuring tissue deformation was predominantly conducted using two methods: imaging of deformation of internal tissues and measurements through external electronic sensors that measured total tissue displacement without measuring internal deformation, such as linear potentiometers or digital calipers. Imaging techniques, such as MRI and US were used with digital image correlation to generate strain or stretch fields in the planes of images [24,43,85,90,114]. Other studies used MRI or US to detect the proliferation of shear waves in tissue during magnetic resonance elastography or strain fields using ultrasound [18,33,45–47, 54,75,76,86,88,91,93,94,96,102,104]. Electronic sensor measurements

of displacement that did not measure internal tissue deformation were not able to generate a strain or stretch field, but several studies used indentation to determine the displacement of tissues and estimate material properties based on the displacement values [19,22,23,34,39,95–97,103]. Sixteen *ex vivo* studies tested in tension [24,55,72,74,77,79–83,107,111–113,115,116], while six tested in compression [71,78,105,106,108,117]. While the most studies used methodologies to collect deformation, data included imaging or electronic sensor measurements, other approaches were used as well. Geometric relationships and laser displacement were used to determine strain in tissue [41,53]. In some cases, strain rate was used in animal and human tissue *ex vivo* [84,107].

4. Discussion

Our objective for this paper was to assemble a collection of studies that investigated the mechanical properties of soft tissues and categorize them by their inclusion of *ex vivo* or *in vivo* tissue samples from humans or animals, the material model used to describe tissue behavior, and the deformation measurements used during experiments. This search resulted in an extensive list of studies reporting material properties of

Table 6
Relationships used to determine material parameters.

No.	Constitutive Law	Relationship
1	Torque – Linear Elastic	$E = \frac{M}{2\pi \cdot 0.4e \cdot R_1 \cdot R_2 \theta}$
2	Elastic	$E_S^{indentation} = \frac{(1-\nu)^2}{\pi d\delta} F_{peak}^{indentaiton}$
3	Modified Elastic	$\mu = f_{contractile} + g_{passive} = \left(\alpha_0 \frac{T}{V}\right) + \left(\frac{k'T}{A} + C\right)$
4	Linear elastic with a statistical recruitment of fibers	$f_n = k \sum_{i=1}^{n} \Delta x_i \sum_{j=1}^{i} \left[\frac{R(x_j)}{(l_0 + x_j)}\right] \Delta x_j$ where $R(x_i) = \frac{e^{\left[-\frac{(x_i-\mu)^2}{2\sigma^2}\right]}}{\sigma\sqrt{2\pi}}$
5	Linear Viscoelastic (1)	$H(t) = H_1 e^{-\frac{t}{\tau_1}} + H_2 e^{-\frac{t}{\tau_2}} + H_\infty$; G(t)=H(t)
6	Linear Viscoelastic (2)	$F(t) = A_1 e^{-k_1 t} + A_2 + A_3 e^{-k_2 t}$
7	Linear Viscoelastic (3)	$\gamma(t) = \gamma_0 \sin(\omega t)$ $\tau(t) = G_d \gamma_0 \sin(\omega t + \delta)$ $G_d = \sqrt{G'^2 + G''^2}$ $\tan(\delta) = \frac{G''}{G'}$
8	Linear Viscoelastic (4)	$G(t) = \frac{3P(t)}{16\delta(t)\sqrt{R\delta(t)}}$ $G(t) = (G_\infty - G_i)\left(1 - e^{-\frac{t}{\tau}}\right) + G_i$
9	Quasi-Linear Viscoelastic	$F(t) = \int_0^t G_t(t-\tau) \frac{\partial F(\delta)}{\partial \delta} \frac{\partial \delta}{\partial \tau} d\tau$ $G_t(t) = \frac{1 + C\left[E\left(t/\tau_2\right) - E\left(t/\tau_1\right)\right]}{1 + C \ln\left(\tau_2/\tau_1\right)}$ * E is the exponential function
10	Power-Law	$\sigma = (k(1-q)E)^{\frac{1}{1-q}}$
11	Voigt	$V_\varphi(\omega) = \sqrt{\frac{2(\mu^2 + \omega^2\eta^2)}{\rho\left(\mu + \sqrt{\mu^2 + \omega^2\eta^2}\right)}}$ $\mu = G = \frac{E}{2(1+\nu)}$

		$\lambda = \frac{Ev}{(1+v)(1-2v)}$
12	**Voigt (2)**	$G_M(\omega) = \mu + i\omega\eta$
13	**Voigt (3)**	$T_{jk} = 2\left(\mu + \eta\frac{\partial}{\partial t}\right)\varepsilon_{jk} + \left(\lambda + \kappa\frac{\partial}{\partial t}\right)\varepsilon_{mm}\delta_{jk}$
14	**Kelvin-Voigt**	$T = -\frac{\omega^2}{\sqrt{(\omega_0^2-\omega^2)^2+4\alpha^2\omega^2}}e^{-i\left(\arctan\left(\frac{2\alpha\omega}{\omega_0^2-\omega^2}\right)\right)}+1$ $R = -\frac{\omega^2}{\sqrt{(\omega_0^2-\omega^2)^2+4\alpha^2\omega^2}}\cos\left[arctan\left(\frac{2\alpha\omega}{\omega_0^2-\omega^2}\right)\right]+1$ $I = -\frac{\omega^2}{\sqrt{(\omega_0^2-\omega^2)^2+4\alpha^2\omega^2}}\sin\left[arctan\left(\frac{2\alpha\omega}{\omega_0^2-\omega^2}\right)\right]$ $S = M\omega^2\left[1-\frac{R-1}{(R-1)^2+I^2}\right]$ $D = \frac{IM\omega}{(R-1)^2+I^2}$
15	**Maxwell**	$G_M(\omega) = \frac{i\omega\eta\mu}{\mu+i\omega\eta}$
16	**Zener**	$G_M(\omega) = \frac{\mu_1\mu_2 + i\omega\eta(\mu_1+\mu_2)}{\mu_2+i\omega\eta}$
17	**Springpot**	$G_M(\omega) = \mu^{1-\alpha}\eta^{\alpha}(i\omega)^{\alpha}$
18	**Friction (1)**	$\mu = \frac{F}{L}$
19	**Friction (2)**	$\mu = \frac{F_{fr}}{F_N}$
20	**Polynomial Stress Relationship**	$\sigma_{muscle} = 3707\lambda_{muscle}^3 - 12117\lambda_{muscle}^2 + 13261\lambda_{muscle} - 4850$ $\sigma_{fat}(kPa) = 2300\lambda_{fat}^3 - 7470\lambda_{fat}^2 + 8160\lambda_{fat} - 2990$ $\sigma_{skin}(kPa) = 2650\lambda_{skin}^2 - 4870\lambda_{skin} + 2227$
21	**Linear Elastic (as a function of shear wave speed)**	$E = \frac{\mu(3\lambda+2\mu)}{\lambda+\mu}$; $E = 3\rho V_s^2$ *where* $V_s = \sqrt{\frac{\mu}{\rho}}$ λ.,μ= Lame´ coefficients in isotropic elastic media
22	**Reduced Elastic Modulus**	$E^* = \frac{\sqrt{\pi}}{2}\frac{K_N}{\sqrt{A}}$; $\frac{1}{E^*} = \frac{1-\nu_1^2}{E_1} + \frac{1-\nu_2^2}{E_2}$ $G^* = \frac{\sqrt{\pi}}{2}\frac{K_N}{\sqrt{A}}$; $\frac{1}{G^*} = \frac{2-\nu_1}{G_1} + \frac{2-\nu_2}{G_2}$
23	**Bec/Tonck**	$E^* = \frac{\pi}{2}\frac{k_z}{\delta}\tan\left(\frac{\pi}{2}-\alpha\right)$; $\frac{1}{k_z} = \frac{e}{f_1(a)\pi a^2 E_f^*} + \frac{1}{f_2(a)2aE_S^*}$;
24	**Song/Pharr**	$G = \frac{(1-\nu^2)E^*}{2(1+\nu)}$; $0.35<\nu<0.5$ E_S^* & E_f^* are respectively the reducedYoung's modulus of thesubstrate (muscle) and later
25	**Perriot/Barthel**	$E^* = E_f^* + \frac{E_S^*+E_f^*}{1+(e_{x0}/a)^n}$; $\beta = \frac{E_S^*}{E_f^*}$
26	**Hugoniot Relationship**	$\dot{s} = C_0 + SV^-$; $\dot{s}$ = Lagrangian shock speed, V^- = particle speed, C_0 = velocity coefficient

soft tissues, which are key for the development of human body models.

The studies reported that mechanical properties of *in vivo* soft tissue, especially in the buttocks and thighs, changed depending on factors such as the body position and the region of the tissues tested [23,25]. These findings indicated that the position and the region of the body, whether or not persons with disabilities were included, and mechanism of deformation measurement need to be considered. For example, if a model of the buttocks in the seated position is desired, then material

properties from the buttocks in the seated position should be used. Material properties from another body position (e.g. prone) or body region (e.g. the leg) are likely to lead to unrealistic predictions of tissue stresses for the buttocks. The magnitudes of the inaccuracies may vary, as the values of material parameters of muscle, fat and skin differed by factors up to two depending on differences in test conditions, body position and body region [48,49,88,95,96,100]. For example, reported shear modulus values for the gastrocnemius varied between 19 and 112 kPa depending on the angle of ankle dorsiflexion. The shear modulus of human muscle and skin ranged from 1 kPa to over 100 kPa [33,46,47,86], and the Young's moduli of skin *in vivo* similarly varied from 8 kPa to 850 kPa [95,98,100]. Further, the elastic modulus of muscle varied from 5 kPa to 126 kPa when using the Voigt model to describe it [42,88]. These large ranges of properties were evidence that factors such as body region, body position, participants with disabilities or able bodied, and the methods of collection all played a role in the material properties reported from *in vivo* tissues.

Ex vivo data also demonstrated large differences from *in vivo* data, which would affect model outputs. Such an example of this is human skin of the forearm, where the *ex vivo* measurements of the Young's modulus were at least 10 times larger than the *in vivo* measurement in the reported studies [24,39,82,95,98,100]. The Young's moduli of *in vivo* human skin samples ranged from 8 to 850 kPa [95,98,100], and *ex vivo* was reported as 0.33–99 MPa (330–99,000 kPa) [24,82]. Samples from *in vivo* and *ex vivo* muscles also varied, as reported values were 3.1–126 kPa [42,88] and 12–129 kPa [71,76], respectively. These large ranges suggest that using the material properties of *ex vivo* tissues as opposed to *in vivo* tissues will significantly alter the results of a model, underscoring the need for material properties that represent *in vivo* human tissue when modeling humans.

Due to variability in experimental conditions and the numerous material models used, there were ranges of material parameters reported [43,53,87,91]. Most studies reported a compressibility parameter (such as D or K), a shear modulus (G or μ), a bulk modulus (κ), a stress tensor (σ), a Young's modulus (E), or material constants (C_1, C_2, etc.). Our review found 47 distinct relationships used to determine the material properties of soft tissues, indicating the breadth of the different models used. Even if the same material model was used, *in vivo* and *ex vivo* studies sometimes reported different parameters as well, making comparison difficult, if not impossible. Studies of the *in vivo* human bicep muscle reported bulk modulus and Young's Modulus, while some *ex vivo* studies reported ultimate tensile strength values, which with the data provided cannot be compared. These differences in models used and parameters reported can make comparisons of material properties between studies difficult if not impossible.

The type of material model to represent each tissue determined the types of material parameters reported; and there were several classes of models used. Types included elastic, hyperelastic, and viscoelastic material models. Model choice is crucial to the description of the soft tissue's mechanical behavior because it affects all computational predictions that use these data. While several models may be appropriate to describe a tissue within the parameters of an experiment or study, models may need to make predictions outside those parameters. It is therefore important to consider that the hyperelastic and viscoelastic models, commonly used to represent muscle, fat, and bulk tissue, will yield different extrapolated tissue responses as compared to linear models. Of the hyperelastic and viscoelastic models chosen to represent muscle, fat, and bulk tissue, the Neo-Hookean and Ogden models were the most common. Furthermore, recognizing the model choice is important because there is no readily available method to convert the material parameters from one model to another. A final consideration for material model choice is the dependence of the model's behavior on pre-strain and residual stresses and strains in the tissue. This is one area currently being studied that needs to be resolved before material properties can be compared between experiments. A method for the consistent determination of a stress-free reference state will eliminate a barrier between comparing experimental results.

Much progress has been made in the characterization of soft tissue, and there is a need to continue improving in this area to better represent human soft tissues. Before material properties of human soft tissue were available, properties of animal tissues were used. However, animal studies have become less frequent as data on human tissues have become more available, and because evidence shows that the properties of the muscle, skin, and fat tissue differ between humans and animals [5,20,40,53,90,96]. As experiments progressed to humans, early properties were determined in *ex vivo* experiments, and those properties were not representative of *in vivo* tissue. More recent tests have begun to collect compressive tissue response data on *in vivo* humans, providing more appropriate data for modelers to use as compared to tensile tests. Finally, there is a growing effort to provide patient-specific soft tissue characterizations, obtained in clinic, to reduce uncertainties related to using population-based data in models [12–16]. These advances will lead to more realistic models of stress/strain predictions and interactions between the human body and its environment. The convergence between the results of any human body model and experimental measurements is integral to establish the credibility of the model. Once the model has been validated by experimental measurements, it can be used to predict behavior and determine how marginal changes in any parameter would affect potential real-life outcomes [121,122].

One tissue that was not documented in this review was fascia. Though fascia does play an important part in the contact between and transduction of forces through other soft tissues, its use in human body models has not been widespread. The exclusion of fascia has been noted as a limitation in some models, as the modelers used simplified contact relationships between skin, muscle, fat, and bone, such as tying the contact nodes together. It is worth repeating that these tissues slide across each other *in vivo*, and as such, this assumption likely affects predicted tissue stresses (i.e. by overestimating interface stresses between tissues). Some models have attempted to consider fascia by including one-layer thick membranes, however its relative thinness when compared to the other tissues, such as skin, muscle, and fat has made it difficult to implement, and even to study, its properties [123]. Several studies have investigated the properties of fascia *in vivo* and *ex vivo*, yet a stated limitation of the *in vivo* studies was that the geometry of the fascia was difficult to model, and thus three-dimensional properties like the Young's modulus have only recently been investigated using some of the methods described in this review [124–128]. The study and inclusion of fascial properties in future literature will further explain their role in force transduction in the body.

Lastly, one area that needs to be explored further is the difference between the material properties of able-bodies individuals and individuals with disabilities. One study investigated the difference in the shear properties of muscle in the legs of individuals with and without disabilities, and two investigated the differences between the soft tissue properties in the buttocks and thighs of able-bodied people and people with spinal-cord injuries [23,129,130]. However, the soft tissue material properties of other regions of the body have not been explored in individuals with disabilities. Persons who use wheelchairs are at an elevated risk of developing PIs, and many wheelchair users have disabilities, therefore it is imperative to expand the knowledge on the material properties of persons with disabilities.

Substantial effort has been dedicated over the last several decades to developing models of the human body for multiple purposes, including PI risk evaluation. However, these models have used a wide range of material properties to represent soft tissues based on a variety of experimental conditions. Recent work has suggested that experimental conditions, such as body region, body position, and whether or not a person has a disability affects the mechanical properties derived for soft tissue [25,31,34,35]. To ensure that human body models most accurately represent the real-life situation they are simulating, the assembly of data presented here show that it is essential to use material properties from analogous experimental conditions.

5. Conclusions

Over several decades, there have been advancements in the material properties used to describe soft tissue for the purposes of human body modeling and for practical applications. These advancements come in the form of determining tissue properties appropriate for wider scopes of applications, starting with *ex vivo* animal tissue properties, up to current experiments describing *in vivo* human tissue properties. Improvements have also been made in the models used to describe the experimental results, whereas early models used linear elastic models for tissue, current descriptions often include hyperelastic or viscoelastic components that capture the strain- or time-dependent properties of tissue. While there has been significant progress in the scope and models of soft tissue properties, there remain several factors that need to be explored further. Among these are body regions of the described tissue, the body position during experiments and its effect on apparent tissue properties, and the effects of physical disability, providing several avenues for future research. Further work will also evaluate current and future experimental methodologies used to deform tissue and measure the deformation.

Declaration of competing interest

There are no competing interests from any of the authors.

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References

[1] Cox SL, Mithraratne K, Smith NP. An anatomically based finite element model of the lower limbs in the seated posture. In: Conference proceedings : . Annual international conference of the IEEE engineering in medicine and biology society IEEE engineering in medicine and biology society conference; 2007. p. 6327–30.

[2] Kiapour A, Kiapour AM, Kaul V, Quatman CE, Wordeman SC, Hewett TE. Finite element model of the knee for investigation of injury mechanisms: development and validation. J Biomech Eng 2014;136:1–14. https://doi.org/10.1115/1.4025692.

[3] Savonnet L, Wang X, Duprey S. Finite element models of the thigh-buttock complex for assessing static sitting discomfort and pressure sore risk: a literature review. Comput Methods Biomech Biomed Eng 2018;21:379–88. https://doi.org/10.1080/10255842.2018.1466117.

[4] Grujicic M, Pandurangan B, Arakere G, Bell WC, He T, Xie X. Seat-cushion and soft-tissue material modeling and a finite element investigation of the seating comfort for passenger-vehicle occupants. Mater Des 2009;30:4273–85. https://doi.org/10.1016/j.matdes.2009.04.028.

[5] Loerakker S, Solis LR, Bader DL, Baaijens FPT, Mushahwar VK, Oomens CWJ. How does muscle stiffness affect the internal deformations within the soft tissue layers of the buttocks under constant loading? Comput Methods Biomech Biomed Eng 2013;16:520–9. https://doi.org/10.1080/10255842.2011.627682.

[6] Luboz V, Petrizelli M, Bucki M, Diot B, Vuillerme N, Payan Y. Biomechanical modeling to prevent ischial pressure ulcers. J Biomech 2014;47:2231–6. https://doi.org/10.1016/j.jbiomech.2014.05.004.

[7] Macron A, Pillet H, Doridam J, Rivals I, Sadeghinia MJ, Verney A, et al. Is a simplified Finite Element model of the gluteus region able to capture the mechanical response of the internal soft tissues under compression? Clin BioMech 2020;71:92–100. https://doi.org/10.1016/j.clinbiomech.2019.10.005.

[8] Siparsky PN, Kirkendall DT, Garrett WE. Muscle changes in aging: understanding sarcopenia. Sport Health 2014;6:36–40. https://doi.org/10.1177/1941738113502296.

[9] Whitney DG, Singh H, Miller F, Barbe MF, Slade JM, Pohlig RT, et al. Cortical bone deficit and fat infiltration of bone marrow and skeletal muscle in ambulatory children with mild spastic cerebral palsy. Bone 2017;94:90–7. https://doi.org/10.1016/j.bone.2016.10.005.

[10] Bouten CV, Oomens CW, Baaijens FP, Bader DL. The etiology of pressure ulcers: skin deep or muscle bound? Arch Phys Med Rehabil 2003;84:616–9. https://doi.org/10.1053/apmr.2003.50038.

[11] Traa WA, van Turnhout MC, Moerman KM, Nelissen JL, Nederveen AJ, Strijkers GJ, et al. MRI based 3D finite element modelling to investigate deep tissue injury. Comput Methods Biomech Biomed Eng 2018;21:760–9. https://doi.org/10.1080/10255842.2018.1517868.

[12] Solav D, Moerman KM, Jaeger AM, Herr HM. A framework for measuring the time-varying shape and full-field deformation of residual limbs using 3-D digital image correlation. IEEE Trans Biomed Eng 2019;66:2740–52. https://doi.org/10.1109/TBME.2019.2895283.

[13] Fougeron N, Rohan PY, Haering D, Rose JL, Bonnet X, Pillet H. Combining freehand ultrasound-based indentation and inverse finite element modeling for the identification of hyperelastic material properties of thigh soft tissues. J Biomech Eng 2020;142:1–8. https://doi.org/10.1115/1.4046444.

[14] Frauziols F, Chassagne F, Badel P, Navarro L, Molimard J, Curt N, et al. In vivo identification of the passive mechanical properties of deep soft tissues in the human leg. Strain 2016;52:400–11. https://doi.org/10.1111/str.12204.

[15] Kappert KDR, Connesson N, Elahi SA, Boonstra S, Balm AJM, van der Heijden F, et al. In-vivo tongue stiffness measured by aspiration: resting vs general anesthesia. J Biomech 2021;114:110147. https://doi.org/10.1016/j.jbiomech.2020.110147.

[16] Li H, Fle´ G, Bhatt M, Qu Z, Ghazavi S, Yazdani L, et al. Viscoelasticity imaging of biological tissues and single cells using shear wave propagation. Front Physiol 2021;9:1–39. https://doi.org/10.3389/fphy.2021.666192.

[17] Blank J, Blomquist M, Arant L, Cone S, Roth J. Characterizing musculoskeletal tissue mechanics based on shear wave propagation: a systematic review of current methods and reported measurements. Ann Biomed Eng 2022;50:751–68. https://doi.org/10.1007/s10439-022-02935-y.

[18] Ceelen KK, Stekelenburg A, Mulders JLJ, Strijkers GJ, Baaijens FPT, Nicolay K, et al. Validation of a numerical model of skeletal muscle compression with MR tagging: a contribution to pressure ulcer research. J Biomech Eng 2008;130. https://doi.org/10.1115/1.2987877.

[19] Linder-Ganz E. Mechanical compression-induced pressure sores in rat hindlimb: muscle stiffness, histology, and computational models. J Appl Physiol 2004;96:2034–49. https://doi.org/10.1152/japplphysiol.00888.2003.

[20] Bosboom EMH, Hesselink MKC, Oomens CWJ, Bouten CVC, Drost MR, Baaijens FPT. Passive transverse mechanical properties of skeletal muscle under in vivo compression. J Biomech 2001;34:1365–8. https://doi.org/10.1016/S0021-9290(01)00083-5.

[21] Lim KH, Chew CM, Chen PCY, Jeyapalina S, Ho HN, Rappel JK, et al. New extensometer to measure in vivo uniaxial mechanical properties of human skin. J Biomech 2008;41:931–6. https://doi.org/10.1016/j.jbiomech.2008.01.004.

[22] Palevski A, Glaich I, Portnoy S, Linder-Ganz E, Gefen A. Stress relaxation of porcine gluteus muscle subjected to sudden transverse deformation as related to pressure sore modeling. J Biomech Eng 2006;128:782–7. https://doi.org/10.1115/1.2264395.

[23] Scott J, Chen S, Roccabianca S, Bush TR. The effects of body position on the material properties of soft tissue in the human thigh. J Mech Behav Biomed Mater 2020:110. https://doi.org/10.1016/j.jmbbm.2020.103964.

[24] Gallagher AJ, Annaidh AN, Bruyere K. Dynamic tensile properties of human skin. In: 2012 IRCOBI conference proceedings; 2012.

[25] Makhsous M, Lin F, Cichowski A, Cheng I, Fasanati C, Grant T, et al. Use of MRI images to measure tissue thickness over the ischial tuberosity at different hip flexion. Clin Anat 2011;24:638–45. https://doi.org/10.1002/ca.21119.

[26] Slade JM, Bickel ÆCS, Modlesky ÆCM, Majumdar S, Dudley ÆGA. Trabecular bone is more deteriorated in spinal cord injured versus estrogen-free postmenopausal women. Osteoporos Int 2005;16:263–72. https://doi.org/10.1007/s00198-004-1665-7.

[27] Modlesky CM, Bickel CS, Slade JM, Meyer RA, Cureton KJ, Dudley GA, et al. Assessment of skeletal muscle mass in men with spinal cord injury using dual-energy X-ray absorptiometry and magnetic resonance imaging. Journal of Applied 2004;96:561–5.

[28] Mahoney ET, Bickel CS, Elder C, Black C, Slade Jr JMDA, et al. Changes in skeletal muscle size and glucose tolerance with electrically stimulated resistance training in subjects with chronic spinal cord injury. Arch Phys Med Rehabil 2005;86. https://doi.org/10.1016/j.apmr.2004.12.021. 2004–6.

[29] Castro MJ, Apple DF, Hillegass EA, Dudley GA. Influence of complete spinal cord injury on skeletal muscle cross-sectional area within the first 6 months of injury. Eur J Appl Physiol Occup Physiol 1999;80:373–8. https://doi.org/10.1007/s004210050606.

[30] Shah PK, Stevens JE, Gregory CM, Pathare NC, Jayaraman A, Bickel SC, et al. Lower-Extremity muscle cross-sectional area after incomplete spinal cord injury. Arch Phys Med Rehabil 2006;87:772–8. https://doi.org/10.1016/j.apmr.2006.02.028.

[31] Moore CD, Craven BC, Thabane L, Laing AC, Frank-Wilson AW, Kontulainen SA, et al. Lower-extremity muscle atrophy and fat infiltration after chronic spinal cord injury. J Musculoskelet Neuronal Interact 2015;15:32–41.

[32] Gorgey AS, Dudley GA. Skeletal muscle atrophy and increased intramuscular fat after incomplete spinal cord injury. Spinal Cord 2007;45:304–9. https://doi.org/10.1038/sj.sc.3101968.

[33] Basford JR, Jenkyn TR, An K-N, Ehman RL, Heers G, Kaufman KR. Evaluation of healthy and diseased muscle with magnetic resonance elastography. Arch Phys Med Rehabil 2002;83:1530–6. https://doi.org/10.1053/apmr.2002.35472.

[34] Scott J, Sheridan B, Andrus R, Monday N, Selby A, Reid Bush T. Tissue matters: in-vivo tissue properties of persons with spinal cord injuries to inform clinical models for pressure ulcer prevention. J Biomech 2021;120:110389. https://doi.org/10.1016/j.jbiomech.2021.110389.

[35] Carda S, Invernizzi M, Cisari C. Sarcopenia or muscle modifications in neurologic diseases: a Lexical or Pathophysiological Difference?, 147; 2012. p. 135–40. https://doi.org/10.13140/RG.2.1.5129.4169.

[36] Lee SH, Park JS, Jung BK, Lee SA. Effects of different seat cushions on interface pressure distribution: a pilot study. J Phys Ther Sci 2016;28:227–30. https://doi.org/10.1589/jpts.28.227.

[37] Chen Y, Wang J, Lung CW, Yang TD, Crane BA, Jan YK. Effect of tilt and recline on ischial and coccygeal interface pressures in people with spinal cord injury. Am J Phys Med Rehabil 2014;93:1019–26. https://doi.org/10.1097/PHM.0000000000000225.

[38] Linder-Ganz E, Shabshin N, Itzchak Y, Gefen A. Assessment of mechanical conditions in sub-dermal tissues during sitting: a combined experimental-MRI and finite element approach. J Biomech 2007;40:1443–54. https://doi.org/10.1016/j.jbiomech.2006.06.020.

[39] Pailler-Mattei C, Bec S, Zahouani H. In vivo measurements of the elastic mechanical properties of human skin by indentation tests. Med Eng Phys 2008;30:599–606. https://doi.org/10.1016/j.medengphy.2007.06.011.

[40] Hendriks FM, Brokken D, van Eemeren JTWM, Oomens CWJ, Baaijens FPT, Horsten JBAM. A numerical-experimental method to characterize the non-linear mechanical behavior of human skin. Skin Res Technol 2003;9:274–83. https://doi.org/10.1034/j.1600-0846.2003.00019.x.

[41] Smalls LK, Wickett RR, Visscher MO. Effect of dermal thickness, tissue composition, and body site on skin biomechanical properties. Skin Res Technol 2006;12:43–9.

[42] Gennisson JL, Deffieux T, Mac´e E, Montaldo G, Fink M, Tanter M. Viscoelastic and anisotropic mechanical properties of in vivo muscle tissue assessed by supersonic shear imaging. Ultrasound Med Biol 2010;36:789–801. https://doi.org/10.1016/j.ultrasmedbio.2010.02.013.

[43] Affagard J-S, Bensamoun SF, Feissel P. Development of an inverse approach for the characterization of in vivo mechanical properties of the lower limb muscles. J Biomech Eng 2014;136:111012. https://doi.org/10.1115/1.4028490.

[44] Affagard JS, Feissel P, Bensamoun SF. Measurement of the quadriceps muscle displacement and strain fields with ultrasound and Digital Image Correlation (DIC) techniques. Irbm 2015;36:170–7. https://doi.org/10.1016/j.irbm.2015.02.002.

[45] Levinson SF. Ultrasound propagation in anisotropic soft tissues: the application of linear elastic theory. J Biomech 1987;20. https://doi.org/10.1016/0021-9290(87)90292-2.

[46] Uffmann K, Maderwald S, Ajaj W, Galban CG, Mateiescu S, Quick HH, et al. In vivo elasticity measurements of extremity skeletal muscle with MR elastography. NMR Biomed 2004;17:181–90. https://doi.org/10.1002/nbm.887.

[47] Then C, Menger J, Benderoth G, Alizadeh M, Vogl TJ, Hübner F, et al. A method for a mechanical characterisation of human gluteal tissue. Technol Health Care 2007;15:385–98.

[48] Sadler Z, Scott J, Drost J, Chen S, Roccabianca S, Bush TR. Initial estimation of the in vivo material properties of the seated human buttocks and thighs. Int J Non Lin Mech 2018;107:77–85. https://doi.org/10.1016/j.ijnonlinmec.2018.09.007.

[49] Chen S, Scott J, Bush TR, Roccabianca S. Inverse finite element characterization of the human thigh soft tissue in the seated position. Biomech Model Mechanobiol 2019. https://doi.org/10.1007/s10237-019-01212-7.

[50] Vogl TJ, Then C, Naguib NN, Nour-Eldin NE, Larson M, Zangos S, et al. Mechanical soft tissue property validation in tissue engineering using magnetic resonance imaging experimental research. Acad Radiol 2010;17:1486–91. https://doi.org/10.1016/j.acra.2010.08.010.

[51] Neumann EE, Owings TM, Schimmoeller T, Nagle TF, Colbrunn RW, Landis B, et al. Data descriptor: reference data on thickness and mechanics of tissue layers and anthropometry of musculoskeletal extremities. Sci Data 2018;5. https://doi.org/10.1038/sdata.2018.193.

[52] Maaß H, Kuhnapfel U. Noninvasive measurement of elastic properties of living tissue. European Control Conference. In: Ecc 1999 - conference proceedings; 1999. p. 2465–70. https://doi.org/10.23919/ecc.1999.7099693.

[53] Suzuki R, Ito K, Lee T, Ogihara N. Parameter identification of hyperelastic material properties of the heel pad based on an analytical contact mechanics model of a spherical indentation. J Mech Behav Biomed Mater 2017;65:753–60. https://doi.org/10.1016/j.jmbbm.2016.09.027.

[54] Chakouch MK, Charleux F, Bensamoun SF. Quantifying the elastic property of nine thigh muscles using magnetic resonance elastography. PLoS One 2015;10. https://doi.org/10.1371/journal.pone.0138873.

[55] Martins PALS, Jorge RMN, Ferreira AJM. A comparative study of several material models for prediction of hyperelastic properties: application to silicone-rubber and soft tissues. Strain 2006;42:135–47. https://doi.org/10.1111/j.1475-1305.2006.00257.x.

[56] Grobbel MR, Shavik SM, Darios E, Watts SW, Lee LC, Roccabianca S. Contribution of left ventricular residual stress by myocytes and collagen: existence of inter-constituent mechanical interaction. Biomech Model Mechanobiol 2018;17:985–99. https://doi.org/10.1007/s10237-018-1007-x.

[57] Rausch MK, Kuhl E. On the effect of prestrain and residual stress in thin biological membranes. J Mech Phys Solid 2013;61:1955–69. https://doi.org/10.1016/j.jmps.2013.04.005.

[58] Pierce DM, Fastl TE, Rodriguez-Vila B, Verbrugghe P, Fourneau I, Maleux G, et al. A method for incorporating three-dimensional residual stretches/stresses into patient-specific finite element simulations of arteries. J Mech Behav Biomed Mater 2015;47:147–64. https://doi.org/10.1016/j.jmbbm.2015.03.024.

[59] Rausch MK, Genet M, Humphrey JD. An augmented iterative method for identifying a stress-free reference configuration in image-based biomechanical modeling. J Biomech 2017;58:227–31. https://doi.org/10.1016/j.jbiomech.2017.04.021.

[60] Mîra A, Carton AK, Muller S, Payan Y. Breast biomechanical modeling for compression optimization in digital breast tomosynthesis. Comput Methods Biomech Biomed Eng 2018. https://doi.org/10.1007/978-3-319-59764-5_4. Lecture No:29–35.

[61] Rajagopal V, Chung J-H, Bullivant D, Nielsen PMF, Nash MP. Determining the finite elasticity reference state from a loaded configuration. Int J Numer Methods Eng 2007;72:1434–51. https://doi.org/10.1002/nme.2045.

[62] Oomens CWJ, Bressers OFJT, Bosboom EMH, Bouten CVC, Bader DL. Can loaded interface characteristics influence strain distributions in muscle adjacent to bony prominences? Comput Methods Biomech Biomed Eng 2003;6:171–80. https://doi.org/10.1080/1025584031000121034.

[63] Levy A, Kopplin K, Gefen A. Device-related pressure ulcers from a biomechanical perspective. J Tissue Viability 2017;26:57–68. https://doi.org/10.1016/j.jtv.2016.02.002.

[64] Makhsous M, Lim D, Hendrix R, Bankard J, Rymer WZ, Lin F. Finite element analysis for evaluation of pressure ulcer on the buttock: development and validation. IEEE Trans Neural Syst Rehabil Eng 2007;15:517–25. https://doi.org/10.1109/TNSRE.2007.906967.

[65] Erdemir A, Guess TM, Halloran J, Tadepalli SC, Morrison TM. Considerations for reporting finite element analysis studies in biomechanics. J Biomech 2012;45:625–33. https://doi.org/10.1016/j.jbiomech.2011.11.038.

[66] Burkhart TA, Andrews DM, Dunning CE. Finite element modeling mesh quality, energy balance and validation methods: a review with recommendations associated with the modeling of bone tissue. J Biomech 2013;46:1477–88. https://doi.org/10.1016/j.jbiomech.2013.03.022.

[67] Hicks JL, Uchida TK, Seth A, Rajagopal A, Delp SL. Is my model good enough? Best practices for verification and validation of musculoskeletal models and simulations of movement. J Biomech Eng 2015;137:1–24. https://doi.org/10.1115/1.4029304.

[68] Oefner C, Herrmann S, Kebbach M, Lange HE, Kluess D, Woiczinski M. Reporting checklist for verification and validation of finite element analysis in orthopedic and trauma biomechanics. Med Eng Phys 2021;92:25–32. https://doi.org/10.1016/j.medengphy.2021.03.011.

[69] Henninger HB, Reese SP, Anderson AE, Weiss JA. Validation of computational models in biomechanics. Proceedings of the Institute of Mechanical Engineers H 2010;224:801–12.

[70] Anderson AE, Ellis BJ, Weiss JA. Verification, validation and sensitivity studies in computational biomechanics. Comput Methods Biomech Biomed Eng 2007;10:171–84. https://doi.org/10.1080/10255840601160484.

[71] Escobar P, Wittles S, Asfour S, Latta L. Mechanical characteristics of muscle, skin and fat-elastic moduli for finite element modeling of limbs. New Orleans, LA: Proceedings of the Orthopaedic Research Society; 2014. p. 15–8.

[72] Nagle AS, Barker MA, Kleeman SD, Haridas B, Douglas Mast T. Passive biomechanical properties of human cadaveric levator ani muscle at low strains. J Biomech 2014;47:583–6. https://doi.org/10.1016/j.jbiomech.2013.11.033.

[73] Ward SR, Tomiya A, Regev GJ, Thacker BE, Benzl RC, Kim CW, et al. Passive mechanical properties of the lumbar multifidus muscle support its role as a stabilizer. J Biomech 2009;42:1384–9. https://doi.org/10.1016/j.jbiomech.2008.09.042.

[74] Lieber R, Runesson E, Einarsson F, Friden J. Inferior mechanical properties of spastic muscle bundles due to hypertrophic but. Muscle Nerve; 2003. p. 464–71.

[75] Yoshitake Y, Miyamoto N, Taniguchi K, Katayose M, Kanehisa H. The skin acts to maintain muscle shear modulus. Ultrasound Med Biol 2016;42:674–82. https://doi.org/10.1016/j.ultrasmedbio.2015.11.022.

[76] Hatta T, Giambini H, Sukegawa K, Yamanaka Y, Sperling JW, Steinmann SP, et al. Quantified mechanical properties of the deltoid muscle using the shear wave elastography: potential implications for reverse shoulder arthroplasty. PLoS One 2016;11. https://doi.org/10.1371/journal.pone.0155102.

[77] Sommer G, Eder M, Kovacs L, Pathak H, Bonitz L, Mueller C, et al. Multiaxial mechanical properties and constitutive modeling of human adipose tissue: a basis for preoperative simulations in plastic and reconstructive surgery. Acta Biomater 2013;9:9036–48. https://doi.org/10.1016/j.actbio.2013.06.011.

[78] Miller-Young JE, Duncan NA, Baroud G. Material properties of the human calcaneal fat pad in compression: experiment and theory. J Biomech 2002;35:1523–31. https://doi.org/10.1016/S0021-9290(02)00090-8.

[79] Alkhouli N, Mansfield J, Green E, Bell J, Knight B, Liversedge N, et al. The mechanical properties of human adipose tissues and their relationships to the structure and composition of the extracellular matrix. Am J Physiol Endocrinol Metab 2013;305:1427–35. https://doi.org/10.1152/ajpendo.00111.2013.-Adipose.

[80] Groves RB, Coulman SA, Birchall JC, Evans SL. An anisotropic, hyperelastic model for skin: experimental measurements, finite element modelling and identification of parameters for human and murine skin. J Mech Behav Biomed Mater 2013;18:167–80. https://doi.org/10.1016/j.jmbbm.2012.10.021.

[81] Reihsner R, Baloghx B, Menzel EJ. Two-dimensional elastic properties of human skin in terms of an incremental model at the in vivo configuration. Med Eng Phys 1995;17:304–13.

[82] Griffin MF, Leung BC, Premakumar Y, Szarko M, Butler PE. Comparison of the mechanical properties of different skin sites for auricular and nasal reconstruction. Journal of Otolaryngology - Head & Neck Surgery 2017;46.

[83] Dunn MG, Silver FH. Viscoelastic behavior of human connective tissues: relative contribution of viscous and elastic components. Connect Tissue Res 1983;12:59–70. https://doi.org/10.3109/03008208309005612.

[84] Silver FH, Freeman JW, Devore D. Viscoelastic properties of human skin and processed dermis. Skin Res Technol 2001;7:18–23.

[85] Ní Annaidh A, Bruy`ere K, Destrade M, Gilchrist MD, Ott´enio M. Characterization of the anisotropic mechanical properties of excised human skin. J Mech Behav Biomed Mater 2012;5:139–48. https://doi.org/10.1016/j.jmbbm.2011.08.016.

[86] Ikeda N, Otsuka S, Kawanishi Y, Kawakami Y. Effects of instrument-assisted soft tissue mobilization on musculoskeletal properties. Med Sci Sports Exerc 2019;51:2166–72. https://doi.org/10.1249/MSS.0000000000002035.

[87] Chakouch MK, Pouletaut P, Charleux F, Bensamoun SF. Viscoelastic shear properties of in vivo thigh muscles measured by MR elastography. J Magn Reson Imag 2016;43:1423–33. https://doi.org/10.1002/jmri.25105.

[88] Levinson SF, Shinagawa M, Sato T. Sonoelastic determination of human skeletal muscle elasticity. J Biomech 1995;28:1145–54. https://doi.org/10.1016/0021-9290(94)00173-2.

[89] Hoyt K, Castaneda B, Parker KJ. Muscle tissue characterization using quantitative sonoelastography: preliminary results. Ultrason Symp 2007;365–8. https://doi.org/10.1109/ULTSYM.2007.101.

[90] Affagard J-S, Feissel P, Bensamoun SF. Identification of hyperelastic properties of passive thigh muscle under compression with an inverse method from a displacement field measurement. J Biomech 2015;48:4081–6. https://doi.org/10.1016/j.jbiomech.2015.10.007.

[91] Then C, Vogl TJ, Silber G. Method for characterizing viscoelasticity of human gluteal tissue. J Biomech 2012;45:1252–8. https://doi.org/10.1016/j.jbiomech.2012.01.037.

[92] Gennisson J-L, Catheline S, Chaffai S, Fink M. Transient elastography in anisotropic medium: application to the measurement of slow and fast shear wave speeds in muscles. J Acoust Soc Am 2003;114:536–41. https://doi.org/10.1121/1.1579008.

[93] Papazoglou S, Braun J, Hamhaber U, Sack I. Two-dimensional waveform analysis in MR elastography of skeletal muscles. Phys Med Biol 2005;50:1313–25. https://doi.org/10.1088/0031-9155/50/6/018.

[94] Dresner M, Rose G, Rossman P, Muthupillai R, Manduca A, Ehman R. Magnetic resonance elastography of skeletal muscle. J Magn Reson Imag 2001;13:269–76.

[95] Zahouani H, Pailler-Mattei C, Sohm B, Vargiolu R, Cenizo V, Debret R. Characterization of the mechanical properties of a dermal equivalent compared with human skin in vivo by indentation and static friction tests. Skin Res Technol 2009;15:68–76. https://doi.org/10.1111/j.1600-0846.2008.00329.x.

[96] Fabrice C, Ehrlacher A, Mohamed R, Ho Ba, Tho MC, Tran V. The in vivo characterization of the mechanical properties of human skin by MRI and indentation techniques. Comput Methods Biomech Biomed Eng 2007;121–2. https://doi.org/10.1080/10255840701479438.

[97] Asserin J, Zahouani H, Humbert P, Couturaud V, Mougin D. Measurement of the friction coefficient of the human skin in vivo: quantification of the cutaneous smoothness. Colloids Surf B Biointerfaces 2000;19:1–12. https://doi.org/10.1016/S0927-7765(99)00169-1.

[98] Diridollou S, Patat F, Gens F, Vaillant L, Black D, Lagarde JM, et al. In vivo model of the mechanical properties of the human skin under suction. Skin Res Technol 2000;6:214–21. https://doi.org/10.1034/j.1600-0846.2000.006004214.x.

[99] Jemec GBE, Selvaag E, Ågren M, Wulf HC. Measurement of the mechanical properties of skin with ballistometer and suction cup. Skin Res Technol 2001;7:122–6. https://doi.org/10.1034/j.1600-0846.2001.70211.x.

[100] Agache PG, Monneur C, Leveque JL, Rigal J de. Mechanical properties and young's modulus of human skin in vivo. Arch Dermatol Res 1980;232:1–12.

[101] Zhang M, Mak AFT. In vivo friction properties of human skin. Prosthet Orthot Int 1999;23:135–41. https://doi.org/10.3109/03093649909071625.

[102] Neumann EE, Owings TM, Erdemir A. Regional variations of in vivo surface stiffness of soft tissue layers of musculoskeletal extremities. J Biomech 2019;95. https://doi.org/10.1016/j.jbiomech.2019.08.001.

[103] Zhang X, Kinnick RR, Pittelkow MR, Greenleaf JF. Skin viscoelasticity with surface wave method. Ultrason Symp 2008. https://doi.org/10.1109/ULTSYM.2008.0156. 651–3.

[104] Kruse SA, Smith JA, Lawrence AJ, Dresner MA, Manduca A, Greenleaf JF, et al. Tissue characterization using magnetic resonance elastography: preliminary results. Phys Med Biol 2000;45:1579–90. https://doi.org/10.1088/0031-9155/45/6/313.

[105] van Loocke M, Lyons CG, Simms CK. A validated model of passive muscle in compression. J Biomech 2006;39:2999–3009. https://doi.org/10.1016/j.jbiomech.2005.10.016.

[106] Aimedieu P, Mitton D, Faure JP, Denninger L, Lavaste F. Dynamic stiffness and damping of porcine muscle specimens. Med Eng Phys 2003;25:795–9. https://doi.org/10.1016/S1350-4533(03)00103-6.

[107] Morrow DA, Haut Donahue TL, Odegard GM, Kaufman KR. Transversely isotropic tensile material properties of skeletal muscle tissue. J Mech Behav Biomed Mater 2010;3:124–9. https://doi.org/10.1016/j.jmbbm.2009.03.004.

[108] Parker KJ, Huang SR, Musulin RA, Lerner RM. Tissue response to mechanical vibrations for "sonoelasticity imaging. Ultrasound Med Biol 1990;16:241–6. https://doi.org/10.1016/0301-5629(90)90003-U.

[109] Geerligs M, Peters GWM, Ackermans PAJ, Oomens CWJ, Baaijens FPT. Linear viscoelastic behavior of subcutaneous adipose tissue. Biorheology 2008;45:677–88. https://doi.org/10.3233/BIR-2008-0517.

[110] Gefen A, Haberman E. Viscoelastic properties of ovine adipose tissue covering the gluteus muscles. J Biomech Eng 2007;129:924–30. https://doi.org/10.1115/1.2800830.

[111] Zhou B, Xu F, Chen CQ, Lu TJ. Strain rate sensitivity of skin tissue under thermomechanical loading. Phil Trans Math Phys Eng Sci 2010;368:679–90. https://doi.org/10.1098/rsta.2009.0238.

[112] Manan NFA, Mahmud J, Ismail MH. Quantifying the biomechanical properties of bovine skin under uniaxial tension. J Med Bioeng 2013;2:45–8. https://doi.org/10.12720/jomb.2.1.45-48.

[113] Łagan SD, Liber-Kne´c A. Experimental testing and constitutive modeling of the mechanical properties of the swine skin tissue. Acta Bioeng Biomech 2017;19:93–102. https://doi.org/10.5277/ABB-00755-2016-02.

[114] Khatam H, Liu Q, Ravi-Chandar K. Dynamic tensile characterization of pig skin. Acta Mechanica Sinica/Lixue Xuebao 2014;30:125–32. https://doi.org/10.1007/s10409-014-0042-9.

[115] Belkoff SM, Haut RC. A structural model used to evaluate the changing microstructure of maturing rat skin. J Biomech 1991;24:711–20. https://doi.org/10.1016/0021-9290(91)90335-K.

[116] Chao CYL, Ng GYF, Cheung K-K, Zheng Y-P, Wang L-K, Cheing GLY. In vivo and ex vivo approaches to studying the biomechanical properties of healing wounds in rat skin. J Biomech Eng 2013;135. https://doi.org/10.1115/1.4025109.

[117] Shergold OA, Fleck NA, Radford D. The uniaxial stress versus strain response of pig skin and silicone rubber at low and high strain rates. Int J Impact Eng 2006;32:1384–402. https://doi.org/10.1016/j.ijimpeng.2004.11.010.

[118] Gefen A, Gefen N, Linder-Ganz E, Margulies SS. In vivo muscle stiffening under bone compression promotes deep pressure sores. J Biomech Eng 2005;127:512. https://doi.org/10.1115/1.1894386.

[119] Hawkins D, Bey M. Muscle and tendon force-length properties and their interactions in vitro. J Biomech 1997;30:63–70. https://doi.org/10.1016/S0021-9290(96)00094-2.

[120] Best TM, McElhaney J, Garrett WE, Myers BS. Characterization of the passive responses of live skeletal muscle using the quasi-linear theory of viscoelasticity. J Biomech 1994;27:413–9. https://doi.org/10.1016/0021-9290(94)90017-5.

[121] Benemerito I, Modenese L, Montefiori E, Mazz`a C, Viceconti M, Lacroix D, et al. An extended discrete element method for the estimation of contact pressure at the ankle joint during stance phase. Proc Inst Mech Eng H 2020;234:507–16. https://doi.org/10.1177/0954411920905434.

[122] Reggiani B, Cristofolini L, Varini E, Viceconti M. Subject-specific FE model for the prediction of the relative micromotion in a total hip implant: verification and validation. Cleveland, OH: ISB XXth Congress - ASB 29th Annual Meeting; 2004.

[123] Al-Dirini RMA, Reed MP, Hu J, Thewlis D. Development and validation of a high anatomical fidelity FE model for the buttock and thigh of a seated individual. Ann Biomed Eng 2016;44:1–12. https://doi.org/10.1007/s10439-016-1560-3.

[124] The in vivo elastic properties of the plantar fascia during the contact phase of walking n.d.

[125] Kitaoka HB, Ping Luo Z, Growney ES, Lawrence Berglund SJ, An K-N. Material properties of the plantar aponeurosis. 1994.

[126] Otsuka S, Shan X, Kawakami Y. Dependence of muscle and deep fascia stiffness on the contraction levels of the quadriceps: an in vivo supersonic shear-imaging study. J Electromyogr Kinesiol 2019;45:33–40. https://doi.org/10.1016/j.jelekin.2019.02.003.

[127] Wang K, Liu J, Wu J, Qian Z, Ren L, Ren L. Noninvasive in vivo study of the morphology and mechanical properties of plantar fascia based on ultrasound. IEEE Access 2019;7:53641–9. https://doi.org/10.1109/ACCESS.2019.2909409.

[128] Bonaldi L, Giulia Fontanella C, Stecco C, Berardo A. Biomechanics of human Fascia Lata: a review of the design, implementation and effectiveness of its mechanical characterization for tissue engineering applications Statement of Significance. [n.d].

[129] Sonenblum SE, Sprigle SH. Buttock tissue response to loading in men with spinal cord injury. PLoS One 2018;13:1–15. https://doi.org/10.1371/journal.pone.0191868.

[130] Linder-Ganz E, Shabshin N, Itzchak Y, Yizhar Z, Siev-Ner I, Gefen A. Strains and stresses in sub-dermal tissues of the buttocks are greater in paraplegics than in healthy during sitting. J Biomech 2008;41:567–80. https://doi.org/10.1016/j.jbiomech.2007.10.011.