Viscoelastic Properties of Young and Old Human Dermis: A Proposed Molecular Mechanism for Elastic Energy Storage in Collagen and Elastin

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ABSTRACT: We have studied the strain-rate dependency of the viscoelastic mechanical properties of human dermis from young (23-year-old) and old (87-year-old) donors using incremental stress–strain measurements. The elastic spring constant for elastic fibers was found to be strain-rate and age dependent, whereas that for collagen was only age dependent. Fibril lengths were observed to decrease with increased strain rates and age for both elastic and collagen fibers; however, the large decrease in collagen fibril viscosity was hypothesized to be a result of thixotropy that results when neighboring collagen fibrils slide by each other. It is concluded that the elastic spring constant measured for elastic fibers may be higher than previously reported and is consistent with stretching of α-helical segments of elastin into a more extended conformation during the initial part of the elastic stress–strain curve. The decrease in the elastic spring constant with increased age observed is consistent with disruption of the elastic fibers and loss of α-helical structure. The pH dependency of the elastic modulus reported previously for collagen suggests that charge–charge interactions within and between collagen molecules are involved in energy storage during stretching. Elastic energy storage is consistent with the stretching of charged pairs located in flexible regions of the collagen molecule. Shear thinning, or thixotropy of skin, is hypothesized to reflect breakage of bonds that occur between collagen fibrils. It is hypothesized that both collagen and elastin are complex macromolecules that are hybrids of flexible and rigid regions. The flexible regions reversibly store elastic energy during stretching by breakage of secondary bonds. After stretching, the flexible regions become extended and transfer stress to the rigid regions of these molecules. This prevents premature mechanical failure of collagen and elastic fibers in the dermis. J Appl Polym Sci 86: 1978–1985, 2002

Key words: skin; viscoelasticity; collagen; elastic tissue; thixotropy

INTRODUCTION

Skin is a multilayered material composed of an epidermis that is about 0.1 mm in thickness, and a dermis that is between 1 and 4 mm thick.[1] The dermis is in turn made up of papillary and reticular layers that contain collagen and elastic fibers. In the young adult, the collagen in the papillary dermis appears as a feltwork of randomly oriented fine fibers, whereas that in the reticular dermis consists of loosely interwoven, large, wavy, randomly oriented collagen bundles.[2] Increased collagen fiber density is observed with increased age and is reported to reflect a decrease in the spaces between individual bundles.[2] The mean fractional volume of collagen fibers determined from stereological data is reported to be between 66 and 69% for both papillary and reticular dermis for all age groups studied.[3] Fibers composed of type I and type III collagens are found in both the papillary and reticular dermis; the type III to type I ratio is somewhat higher in the papillary layer compared to that in the reticular layer,[4] with type I collagen comprising about 80 to 90% of the total collagen content. Collagen fibers are observed to be more compact with increased age and appear to unravel.[2] Collagen concentration in the skin of rats increases up to 6 months of age, after which it decreases.[5] The type III collagen content of rat skin decreases from 33% (2 weeks) to 18.6% at 1 year.[5]

Elastic fibers are composed of an amorphous core of elastin, which constitutes about 90% of the fiber, and a microfibrillar component consisting of 10- to 12-nm diameter fibrils.[6] Early studies of the retractive force as a function of temperature made on elastic fibers showed the mechanical behavior of elastin was similar to that of a rubber made up of a random coil network.[7] The predominant deformation of this random coil network was hypothesized to be entropic in nature.[8] Later studies conducted on elastin suggested that the alanine-rich sequences in the crosslink region form an α-helix[9] and other sequences with hydrophobic residues form β-spirals.[10]

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Elastic tissue in skin consists of superficial thin bundles of microfibrils that become associated with progressively larger amounts of amorphous elastin and increase in size from the papillary to reticular dermis.[3] The relative volume of elastic fibers increases from about 0.7 to about 2.5%, whereas the diameters increase from about 1 to 2 μm in going from the papillary to the reticular dermis.[3] The volume fraction of elastic tissue increases with age up to about 1 year then appears to plateau. The density of elastic fibers in the papillary dermis decreases from about 2.5 to about 2% for individuals older than 10 years. Elastic fibers from skin from older individuals appear to fray and contain holes.[2]

Skin contains a number of glycosaminoglycans (GAG) including hyaluronan and dermatan sulfate[11]; the hyaluronan content of skin has been estimated to be between 0.03 and 0.09%.[12] The GAG content is reported to decrease with respect to the amount of protein with increased age.[13] Dermatan sulfate proteoglycans from skin and cartilage have been isolated and are similar in structure.[14] They have relatively small molecular sizes and consist of one (decorin) or two (biglycan) GAG chains attached to a core protein with a molecular weight of about 40 kDa.[15] Both decorin and biglycan share homologous core proteins containing 7–24 repeats of characteristic leucine-rich amino acid motifs.[15] Decorin has been shown to bind to type I collagen by leucine-rich regions designated 4–5, whereas biglycan does not bind to type I collagen.[16]

The mechanical properties of skin reflect the behavior of the dermis because removal of the epidermis does not change the viscoelastic properties of skin.[17] The mechanical response of skin to applied loads involves both a viscous component associated with energy dissipation and an elastic component associated with energy storage.[18,19] Dissipation of loads applied to skin occurs by molecular and viscous sliding of collagen fibrils during alignment with the force direction,[20] whereas elastic energy is stored as a result of stretching of flexible regions in the collagen triple helix.[17,21,22]

The stress–strain behavior of skin is composed of three phases.[18] Up to strains of about 0.3 the collagen network offers little resistance to deformation and the behavior is dominated by the elastic fibers.[23] Between strains of about 0.3 and 0.6 the collagen fibrils begin to offer resistance to deformation. During this linear portion of the stress–strain curve, the elastic component dominates the deformation[19] and appears to involve stretching of the flexible regions within crosslinked collagen molecules.[21,22,24] In the tendon, the viscous component of the stress–strain behavior is associated with fibrillar slippage.[21,22,24] The yield and failure region (strains above 0.6 for skin) involves fibril defibrillation.[18]

Results of viscoelastic tests conducted on human dermis from donors between the ages of 47 and 86 years suggest that the elastic spring constant of the collagen molecule in dermis is about 4.4 GPa compared to a value of 8 GPa found for tendon.[17,21,22,24] Collagen fibril lengths reported for skin and dermis are about 50 μm[17] compared to between 400 and 900 μm for tendon.[21,24] The lower value of the elastic spring constant for collagen in skin compared to that found for tendon was hypothesized to be a result of the higher molecular tilt angle reported for skin.[17] The purpose of this study was to analyze the viscoelastic properties of young and old skin to determine the effects of aging and strain rate on the elastic spring constant and collagen fibril length.

EXPERIMENTAL

The mechanical behavior of human dermis was determined from incremental stress–strain tests conducted on processed dermis as previously described.[21,22,24] Skin was procured from tissue banks in compliance with all applicable federal and state regulations and with American Association of Tissue Banks standards. It was cryopreserved for storage before processing using standard skin bank procedures. Frozen skin was removed from storage and then thawed before processing. It was then treated with solutions to remove epidermis and other cellular materials, virally inactivated, and extensively washed to remove residuals of decellularization and viral inactivation solutions.[17] The processed dermis was then frozen and freeze-dried before mechanical testing.

Strips of processed human dermis (50 $\times$ 4 cm cross section by 1–2 mm thick) were hydrated in phosphate buffer solution (PBS) at room temperature for a minimum of 30 min before mechanical testing. Sample dimensions were measured by observation through the calibrated eyepiece of a Leitz Pol microscope (Leitz, Rockville, NJ). The dimensions were measured in three places along the strip and the average cross-sectional area was obtained. It was assumed that the sample cross section was rectangular.

Standard viscoelastic testing was conducted on wet samples at room temperature as previously described.[21,22] Strips were clamped into the Instron grips (Canton, MA) using sandpaper on each side of the specimen, and samples were stretched in tension at strain rates of 10 and 1000% per min using an Instron Model 1122 testing device. The gauge length used was 2 cm. All strains were determined from the displacement of the crosshead. Specimens were subjected to strain increments of 5%, resulting in an incremental stress–strain curve. After each strain increment the stress was allowed to decay to its equilibrium value before an additional strain increment was added. The elastic component of the stress was defined as the

stress at equilibrium, whereas the viscous component was calculated from the difference between the total stress and the elastic component. Total, elastic, and viscous stress–strain curves were approximated by using a polynomial curve-fitting program within Cricket Graph (Malvern, PA). Approximately eight samples were tested at each strain rate.

Lines representing the viscous stress–strain curves obtained from incremental stress–strain curves were converted into fibril lengths based on estimation of axial ratios and shape factors as described previously.[21,22] Viscous stress–strain equations were approximated by straight lines as discussed above and the equations were divided by the strain rate (0.1/min or 10/min) to give an equation that represented the extensional viscosity versus strain in MPa-s. The shear viscosity as a function of strain was then approximated from the extensional viscosity by dividing by a value of 3.0, which is equivalent to the relationship between shear modulus and tensile modulus for isotropic materials with a Poisson's ratio of 0.5. Shear viscosity as a function of strain was converted into shape factor V by dividing by the solvent viscosity (8.23×10^{-4} MPa-s) for water and by dividing by the volume fraction of polymer. The volume fraction of collagen was estimated from the wet and dry sample weights. The axial ratio Z was estimated from the following equation, where k is 0.1395 for collagen and it is assumed that the molecule can be modeled as a prolate ellipsoid:

$$Z = (V/k)^{0.552} \quad (1)$$

Collagen fibril lengths were calculated by multiplying eq. (1) by the estimated value of the average fibril diameter for collagen (80 nm). The elastic fiber diameter was approximated using eq. (1) and the diameter of elastic fibers (2.0 μm) in the reticular layer reported by Smith.[1]

Collagen polymer volume fractions were calculated from the dry and wet weights of skin, assuming that the collagen had a specific gravity of 1.33 after subtracting the volume fraction of elastic fibers (0.02).

Statistical analyses were conducted using Student's t-test and looking for significant differences between data sets. Significant differences are reported only for data sets that were found to be different at 0.95 or greater confidence level.

RESULTS AND DISCUSSION

Elastic and viscous stress–strain curves for human dermis at strain rates of 10 and 1000% per min are shown for a 23-year-old Caucasian male [Fig. 1(a)] and an 87-year-old Caucasian female [Fig. 1(b)]. Both elastic and viscous stress–strain curves are shifted by increasing the strain rate. The slopes of the elastic and viscous stress–strain curves at low and high strains are tabulated in Table I. Analysis of data in Table I indicates that in general the slope of the initial portion of the elastic stress–strain curve increases with increased strain rate as well as the slopes of the initial and final portions of the viscous stress–strain curves. The final slope of the elastic stress–strain curve does not appear to change with strain rate but may decrease with increased age. Elastic spring constants and fibril lengths for elastic and collagen fibers are tabulated for young and old dermis at low and high strain rates in Table II. The initial slope of the elastic stress–strain curve increases from 2.10 to 4.30 MPa for young dermis as the strain rate increases from 10 to 1000% per min, whereas for old dermis the increase is from 0.60 to 1.08 MPa. At high strains the elastic slope increases from 41.0 to 45.0 MPa for young dermis and decreases from 15.0 to 7.31 MPa for old dermis. The initial slope of the viscous stress–strain curve increased from 1.10 to 4.60 MPa for young dermis as the strain increased from 10 to 1000% per min, whereas for old dermis it increased from 0.260 to 1.85 MPa. At high strain, the viscous slope increased from 25.0 to 46.0 MPa as the strain rate increased from 10 to 1000% per min for young dermis, whereas for old dermis it increased from 5.40 to 8.50 MPa.

Collagen volume fractions of young and old skin were calculated from the wet and dry weights, assuming that the specific gravity of the polymer was 1.33 and the volume fraction of elastic fibers was 0.02 as reported previously.[25] This gave volume fractions of 0.166 and 0.096 for collagen in young and old skin, respectively.

Elastic spring constants for the elastic fibers (initial part of the elastic stress–strain curve) and for collagen (upper part of the elastic stress–strain curve) were calculated from the slope of the elastic stress–strain curves after correction for the volume fraction of polymer. The elastic spring constant for elastic fibers increased from 0.105 to 0.214 MPa for young dermis when the strain rate was increased from 10 to 1000% per min, whereas for old dermis the value increased from 0.030 to 0.054 MPa (Table II). The elastic spring constant for collagen was 3.70 MPa at 10%, whereas at 1000% it was 4.05 MPa for young skin. For old dermis the elastic spring constant was 2.50 MPa (10% per min) and 2.04 MPa (1000% per min).

Elastic and collagen fibril lengths were calculated using eq. (1) and by assuming that each macromolecule could be modeled as a prolate ellipsoid with a value of the constant k as 0.1395. These parameters were used previously for collagen[21,22,24] and are only approximations for elastic tissue because solution parameters are not available for elastin or elastic tissue.

Estimated elastic fiber lengths for young skin decreased with increased strain rate from 845 to 443 μm, whereas for old dermis they decreased from 390 to 272

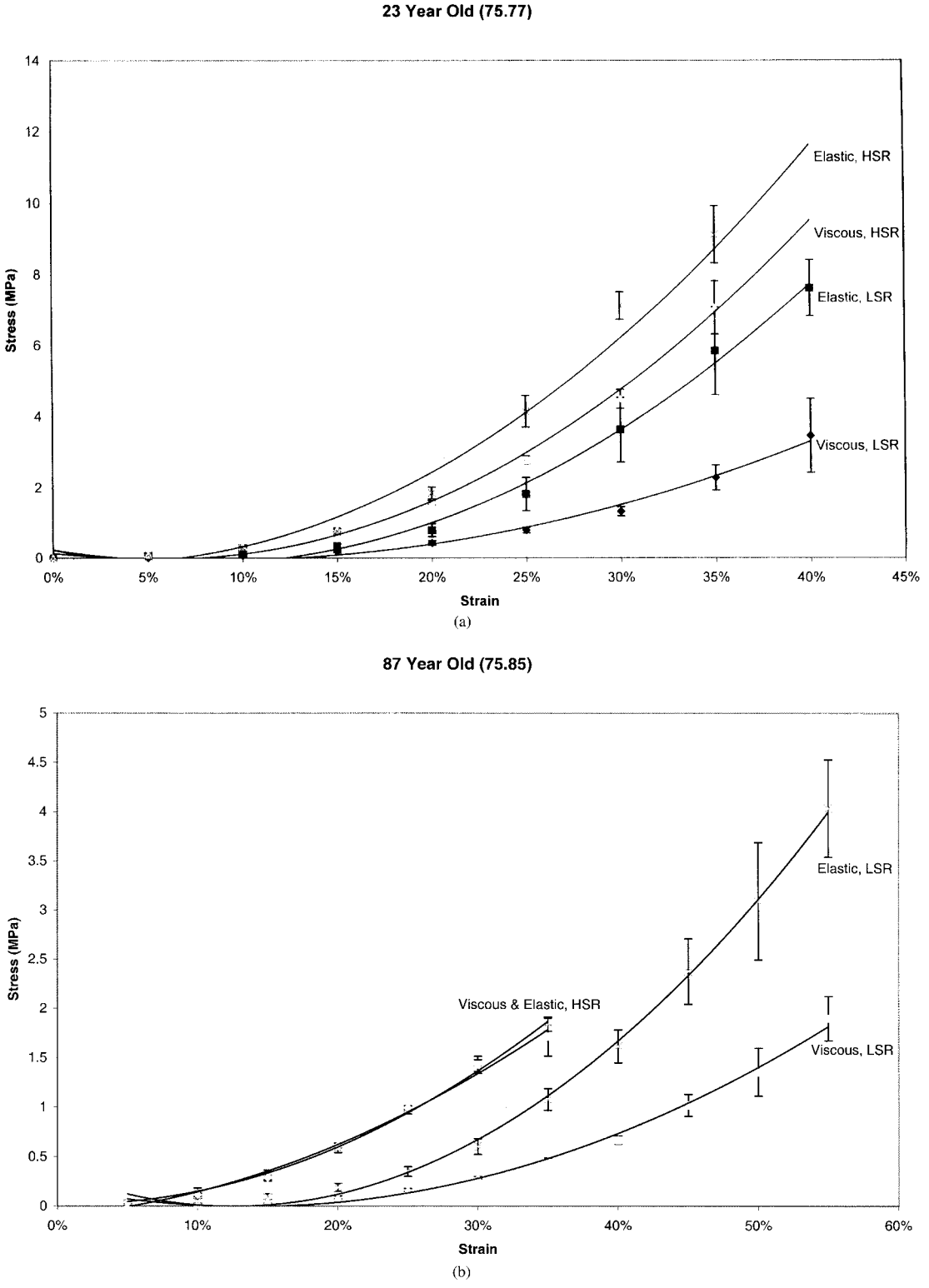

Figure 1 Elastic and viscous stress–strain curves for dermis from (a) a 23-year-old Caucasian male donor and (b) an 87-year-old Caucasian female donor at strain rates of 10% [low strain rate (LSR)] and 1000% [high strain rate (HSR)] per min. Total stress was obtained from the initial stress observed in an incremental stress–strain experiment, whereas elastic stress was obtained from the equilibrium value of the stress at a fixed strain. Viscous stress was defined as the difference between the total and elastic stresses at each strain increment. Error bars shown are standard deviations about the mean values.

TABLE I
Slopes of the Incremental Stress–Strain Curves for Dermis at Strain Rates of 10% and 1000% per min Determined Using a Least-Squares Linear Fit to the Low- and High-Strain Slopes

	Slope (MPa)		Slope (MPa)	
Sample	Initial elastic	Final elastic	Initial viscous	Final viscous
23-year-old male				
10%	2.10	41.0	1.10	25.0
1000%	4.30	45.0	4.60	46.0
87-year-old female				
10%	0.60	15.0	0.26	5.40
1000%	1.08	7.30	1.85	8.50

μm with increased strain rate (Table II). Collagen fibril lengths decreased from 179 to 19.6 μm for young skin as the strain rate was increased, whereas for old dermis it decreased from 103 to 10.4 μm as the strain was increased.

The mechanical behavior of skin has been studied in great detail by a number of investigators.[18,23,26–29] We previously reported that the mechanical properties of skin and processed dermis are not statistically different.[17] In this study we report the strain rate dependency of young and old dermis. The low strain elastic and viscous properties in this study were observed to be strain rate dependent, whereas only the high strain viscous properties were strain rate dependent. The results of these studies suggest that elastin (elastic fibers) plays a role in the mechanical behavior of skin at small stress values and small deformations, whereas collagen resists deformation in the linear, yield, and failure regions.[18,23] However, the molecular basis for the strain-rate dependency of the mechanical properties of skin is not addressed in previous studies.[18,27,29,30,31] Vogel[27] reported for rat skin that the strain to failure was independent of strain rate but the load to failure and tensile strength were positively correlated with the logarithm of strain rate. He later showed in the rat that the ultimate load, tensile strength, and ultimate modulus of elasticity show a sharp increase during maturation (0 to 12 months) and a decrease during aging (12 to 24 months).[28] In contrast, relaxation was shown to decrease during both maturation and aging.[26]

Finlay[29] used a rotational device applied to the skin to apply a torque to the surface. He reported that the applied torque preceded the displacement in time and that the skin showed a reduced viscosity with increased strain rate, characteristic of thixotropic behavior.[29] Potts et al.[32] studied the attenuation of shear waves in human skin at frequencies between 0 and 1000 Hz. They reported that the viscosity of wet skin decreased from about 225 to about 40 dyn-s/cm^2 at frequencies between 0 and 500 Hz.[32] Kronick[30] reported that the dynamic elastic modulus increased with increased strain and was higher at neutral pH compared to that at pH values below 4.0 and above 9.0.

Previously we studied the viscoelastic behavior of human skin and dermis to separate the elastic (time-independent) and viscous (time-dependent) contributions.[17,19] Results of our previous studies suggest that the stress–strain curves of skin and dermis are similar. Total stress–strain curves were separated into elastic and viscous stress–strain curves, and elastic spring constants for elastic fibers and collagen were determined from the initial and final slopes of the elastic stress–strain curves.[17] Values of the elastic spring constants reported previously were 4.0 MPa and 4.4 GPa, respectively, for elastic and collagen fibers. Fibril lengths were approximated for collagen fibers from the final slope of the viscous stress–strain curves. Calculated values of the fibril length for skin and dermis were between 49 and 64 μm.[17]

Results presented in this study suggest that careful analysis of the initial portion of the elastic stress–strain curve indicates that the elastic spring constant for elastic fibers is higher than previously reported and varies from about 0.054 to 0.214 GPa. The increased spring constant reported here compared to that of a previous report[17] arises as a result of measuring the equilibrium force at 5% strain intervals using lower full-scale loads. Previously, we reported equilibrium force measurements starting at a strain of 30% with strain increments of 20%.[17] Increased values of the elastic spring constant at higher strain rates suggest that rapid stretching may prevent conformational changes in elastin molecules that may occur at lower strain rates. At low strain rates for young skin, the elastic spring constant is about 105 MPa and is close to the estimated stiffness of the stratum corneum of skin (120 MPa).[33] The stratum corneum is composed of keratin molecules in a predominantly α-helical structure.[33] This value (120 MPa) is much lower than the estimated stiffness of silk (7 GPa), which is in a β-structure.[33] This would support the hypothesis that α-helical segments in elastin molecules provide the resistance to deformation at low strains in skin. The decrease in the spring constant of elastic fibers at low strain rates may reflect the unraveling of α-helical segments during deformation. This would lower the elastic spring constant, given that the resistance to deformation would involve stretching both α-helices and uncoiled chains. The force required to stretch an uncoiled chain devoid of hydrogen bonding would be less than that required to stretch an α-helical segment. The decreased value of the elastic spring constant that occurs with increased age is consistent with reports of fraying and hole development in the elastic fibers from old skin[2] and a decrease in the elastic fiber length with increased age.

TABLE II
Elastic Spring Constants and Fibril Lengths for Collagen and Elastic Fibers from Young and Old Skin

	Elastic spring constant (GPa)		Fibril length (μm)	
Sample	Elastic fibers[a]	Collagen[b]	Elastic fibers	Collagen
23-year-old male				
10%	0.105	3.70	845	179
1000%	0.214	4.05	443	19.6
87-year-old female				
10%	0.030	2.51	390	103
1000%	0.054	2.04	272	10.4

[a] Volume fraction assumed to be 0.02.
[b] Volume fraction calculated to be 0.166 and 0.096 for young and old skin, respectively.

Recent studies of elastin model peptides suggest that poly(valine–proline–glycine–valine–glycine) forms one β-turn per pentameric unit with proline–glycine at the corner of the bend and a 4-1 hydrogen bond connecting the C═O group of the first valine to the NH of the other valine along the sequence.[34] The repetition of this conformational unit gives rise to a helical arrangement termed the β-spiral.[35] β-Turns present in this structure act as spacers between the turns of the spiral. Between β-turns are found valine–glycine dipeptides that can undergo large rotational changes.[35] In another model,[35] nonrecurring, isolated type II β-turns are proposed for sequences containing glycine–X–glycine–glycine–X. These have X–glycine or glycine–glycine sequences at the corners with 4-1 hydrogen bonds connecting the first and fourth glycine or the second and fifth X residue, respectively. The turns are believed to be labile, giving rise to a sliding β-turn that can absorb energy during stretching.[35] Another sequence, alanine–lysine–glycine–glycine–glycine–alanine–lysine–glycine found at the N-terminal regions of human elastin, was reported to form a polyproline II like helix that is in equilibrium with other conformations.[36] These reports suggest that a number of different extended conformations can be adopted by segments of the elastin molecule. An elastin molecule that is a composite of α-helical, random coil, and extended conformations could store elastic energy during stretching by a transition from α-helical and random coils to one of several different, more extended conformations. A recent report[37] indicates that elastin in rat, bovine, and human tissues contains about 10% α-helical, 40% β-sheet, and about 50% with an undefined structure, which is consistent with the hypothesis that elastic energy is stored during stretching by extending more compact structural elements.

The observed elastic spring constant for collagen is between 3.7 and 4.0 GPa and is slightly smaller than previously reported.[17] However, the decreased elastic spring constant observed with increased age is consistent with fragmentation of the collagen fibers and destruction of the collagen network in skin. These results are consistent with a decreased collagen fibril length associated with aging. Kronick[30] reported that the elastic modulus of collagen was dependent on the pH. At pHs approaching the pK_a of the free carboxylic (3.5) and free amino groups (10.6), the elastic modulus decreased, suggesting that charge–charge interactions within and between collagen molecules provide resistance to elastic stretching. This is consistent with the hypothesis that interactions between charge pairs present on the outside of the collagen triple helix stabilize the molecule[38] and that elastic energy storage is associated with breakage of charged pairs that are within and between collagen molecules.

The apparent decrease in the collagen fibril viscosity and length associated with an increase in the strain rate reflects a decrease in the viscosity of skin. The observation that skin is thixotropic is consistent with results reported by Findlay[29] and Potts et al.[31] Findlay[29] attributed the shear thinning to thixotropy of the GAGs present in skin. This would suggest that interactions between collagen fibrils during deformation might involve PG molecules that interact with the surface of adjacent fibrils. Because decorin is the predominant PG in skin that interacts with collagen type I, it is possible that collagen fibril slippage involves breakage of bonds between decorin and collagen molecules during fibrillar slippage. Previous studies have shown that polysaccharides that form extended structures in solution at low shear rates undergo large decreases in axial ratio at high strain rates, leading to thixotropic behavior.[39] However, the results of recent viscoelastic mechanical studies on self-assembled type I collagen fibers suggest that, in the absence of PGs, the viscosity of collagen fibrils decreases when the strain rate is increased from 10 to 1000% per min.[40] This observation suggests that thixotropy of skin may not be dependent on the presence of PG molecules.

Another important issue relating to energy storage and dissipation in skin involves the type(s) of connections that occur between collagen and elastic fibers. It is generally accepted that elastic fibers bear the loads at low strains, whereas collagen fibers store elastic energy at high strains.[18] The low concentration of elastic fibers in human skin (about 2%) suggests that most of the load that is applied to skin must be supported by collagen fibers. However, the stress–strain curve of rat skin[18] is more like that of an avian ligament,[41] in which the collagen and elastic fibers are interdigitated together. The ligamentum propatagiale runs from the shoulder to wrist in birds' wings and supports the leading edge of a skin fold in this region. The stress–strain curve of this ligament is very similar to that of rat skin, with a maximum stress approaching 5 MPa and a strain at failure of about 140%.[41] This ligament is composed of collagenous end segments that are relatively inextensible, connected by a center

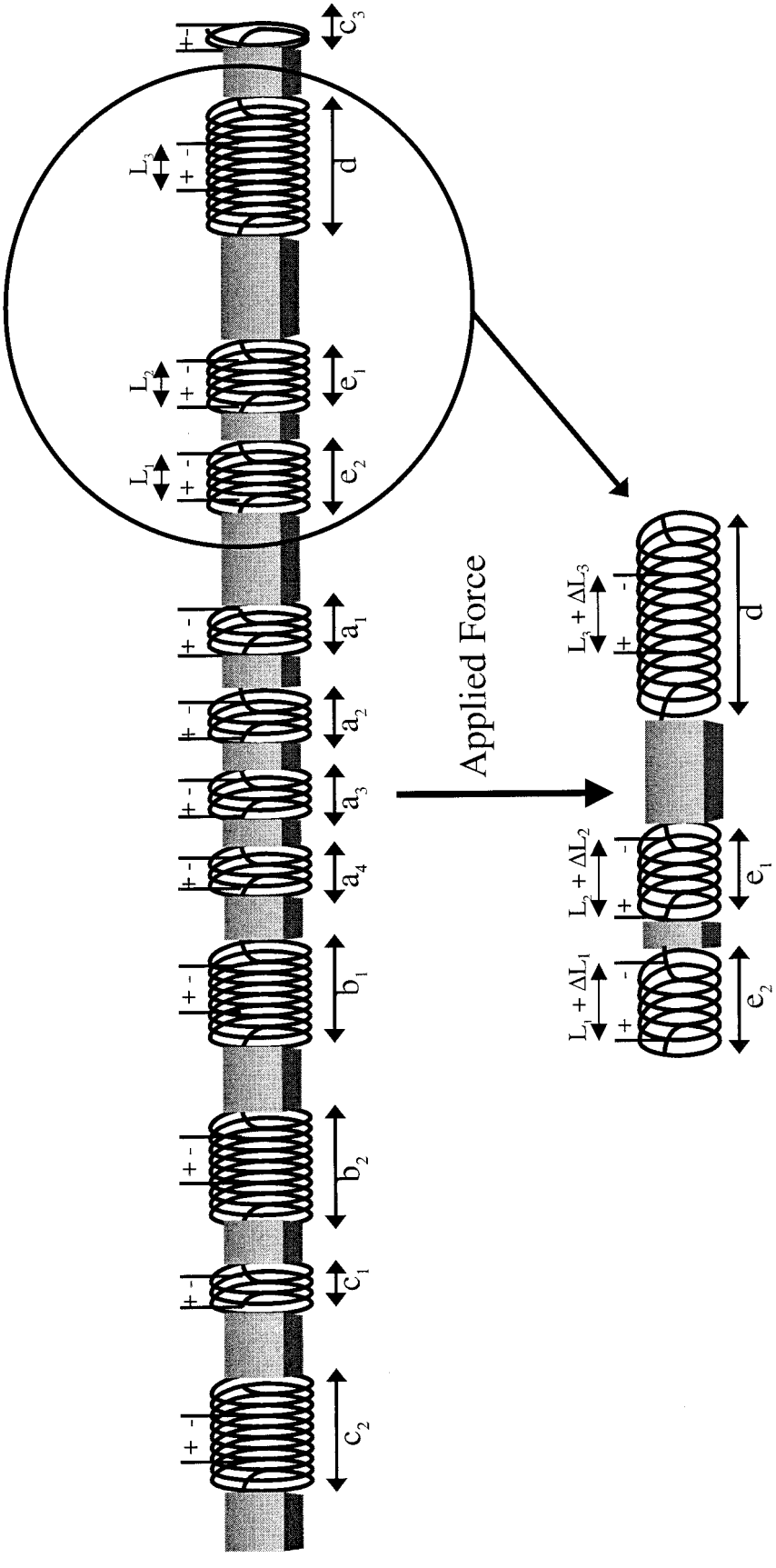

Figure 2 A hypothesized mechanism for elastic energy storage in collagen. The diagram illustrates a collagen molecule comprised of 12 rigid regions (solid areas) separated by 12 flexible regions (springs). During the application of a tensile force to the dermis, the flexible regions (12 positive staining bands of the collagen D period, c_2, c_1, b_2, b_1, a_4, a_3, a_2, a_1, e_2, e_1, d, and c_3) are stretched and the energy is absorbed by breaking electrostatic interactions (+ −) between amino acid side chains that make up the charged pairs within a molecule (shown) and between molecules (not shown).

section that is highly extensible and is composed of interdigitated collagen and elastic fibers.[41] An interesting aspect of the mechanical behavior of this tissue is that the junctions between segments are sufficiently strong so that mechanical failure commonly occurs in the collagenous segments or their attachments to the tissues of origin.[41] Brown and colleagues[41] have modeled this tissue as a parallel arrangement of collagen and elastic fibers with the collagen fibers buckled to allow for extension of elastic fibers at low strains. At high strains the collagen fibers bear the load after the collagen fibers are straightened.[41] If their model is correct, then the strength of the elastic tissue must approach that of the collagenous ends, given that failure would occur in the central sections if the elastic tissue was much weaker than the collagenous tissue. This observation would tend to support the hypothesis that at high extensions, the mechanical behavior of elastic tissue approaches that of collagenous tissue. This in turn supports structural models for elastin at high strains that have extended conformations, such as the β-conformation, given that the stiffness of random conformations and α-helices is much lower than that of collagen and β-conformations.[33]

CONCLUSIONS

Both collagen and elastin are complex molecules that contain rigid and flexible sequences. In type I collagen, the flexible regions are hypothesized to be associated with the 12 positive staining bands in the collagen fibril[42] (see Fig. 2). In elastins regions with a polyproline II like helix in the N-terminal end,[36] α-helical, random coil, and β conformations are present.[34–37] We propose that elastic energy storage in both of these molecules occurs in the flexible regions, whereas the more rigid sections help to transfer stress throughout the molecule.

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