112. Lindgren, L.: Der Biomechanismmus der Zervixdilation wahrend der Geburt, *Gynaekol. Runds. (Basel), 11*, 237–276, 1971.

113. Lindgren, L.: The causes of fetal head moulding in labor, *Acta Obstet. Gynecol. Scand., 39*, 46–62, 1960.

114. Schwarcz, R. L., Strada-Saenz, G., Althabe, O., Fernandez-Funes, J., and Caldeyro-Barcia, R.: Pressure exerted by uterine contractions on the head of the human foetus during labor, in *Perinatal Factors Affecting Human Development*, Scientific Pub. No. 185, Pan American Sanitary Bureau, Washington, 1969, p. 115.

115. Holmlund, D., and Lindgren, L.: Friction between the fetal head and uterine wall during normal labor and lower uterine spasm, *Am. J. Obstet. Gynecol., 103*, 939–941,1969.

116. Rice, D. A., Yang, T. Y., and Stanley, P. E.: A simple model of the human cervix during the first stage of labor, *J. Biomech., 9*, 153–163, 1976.

117. Rice, D. A.: Mechanism and measurement of cervical dilatation, Ph.D. thesis, Purdue University, W. Lafayette, Ind., 1974.

118. Romero-Salinas, G., Pantle, J., Aramburú, G., Figueroa, J. G., Bieniarz, J., and Caldeyro-Barcia, R.: Registro continuo de dilatacion cervical en el parto humano con método electronico directo, *Fourth Uruguayan Cong. Obstet. Gynecol., 2*, 718, 1964.

119. Egan, J. M., and Richardson, J. D.: Biomechanical analysis of the deformation of the lower uterus during labour, in R. Huiskes, D. van Campen, and J. DeWijn (eds.), *Biomechanics: Principles and Applications*, Martinus Nijhoff, The Hague, 1982, pp. 403–409.

120. Richardson, J. D., Sutherland, I. A., Allen, D. W., and Dore, F.: The development of an instrument for monitoring dilatation of the cervix during labour, *Biomed. Eng., 2*, 311–313, 1976.

CHAPTER 11

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Skin Mechanics

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INTRODUCTION

The skin forms a barrier between the body and the external environment. Its functions [1, 2] are to contain and support internal organs, to protect them from unfavorable external stimuli (chemical, thermal, mechanical, biological, and radiative), to participate in maintaining a proper internal milieu, and to serve as a sensory interface with the external world.

Mechanically, the most important functions of the skin are to support internal organs and protect them (from abrasions, blunt impact, cutting, and penetration), while at the same time allowing considerable mobility. The skin's ability to perform this variety of functions results from a unique combination of mechanical properties which are closely related to its microstructure. Both properties and structure change with body site, age, sex, weight, exposure, pregnancy, and disease, and under the influence of drugs and chemicals.

Skin mechanics, apart from being an interesting subject in itself, has considerable clinical impact, especially in plastic surgery. The present chapter will be devoted to the mechanical and structural characteristics of the skin and their significance in plastic surgery.

11.1 CONSISTENCY AND STRUCTURE OF THE SKIN

The skin accounts for about 16 percent of the body weight in human adults. Its surface area is 1.5 to 2.0 m^2 in adults, and it varies in thickness from 0.2 mm (eyelid) to 6.0 mm (sole of foot) [3]. The skin consists of an outer layer, the epidermis, and the dermis. The latter contains collagen, elastin, reticulin, fibrocytes, blood and lymph vessels, nerve endings, hair and hair follicles, and glands (sebaceous, sweat, and apocrine) together with their associated ducts. All are embedded in what is called the *ground substance*—a gelatinous matrix consisting of water, mucopolysaccharides (notably hyaluronic acid and chondroitin sulfate), proteins (mostly soluble collagen), enzymes, and electrolytes.

The mechanical and barrier functions of the epidermis stem primarily from the stratum corneum—an outer layer of keratinized cells. It is from ~10 to 200 μm thick (sole of foot) and is organized as a multicellular membrane in which the cells (~0.8 μm thick) are stacked in columns and interdigitate with cells of adjacent columns [4].

The predominant components in the dermis are the ground substance and collagen fibers. The collagen accounts for 60 to 80 percent of the dry weight of the skin depending on age, sex, and site [5, 12]. The collagen (primarily of type I, with lesser amount of type III) appears as a three-dimensional, apparently disordered, network of wavy or coiled fibers. They are structured in several layers of primarily planar networks [6], with some fibers running between them [7]. Scanning electron microscopic observation [8] suggests that the collagen fibers near the epidermis (papillary layer) are finer and randomly oriented, while at the midzone they are coarser and densely packed, and may show preferred orientation. In the deep zone they are coarse again but are loosely packed.

The ground substance matrix accounts for 70 to 90 percent of the skin's volume [9]. Although it was traditionally considered to be amorphous, there is evidence [10, 11] which suggests that some components of the ground substance are incorporated in the structure of the collagen fiber.

Elastin accounts for about 4 percent of the dry weight [12, 13]. It appears as fine fibers which intertwine around the thicker collagen fibers in the deep layers of the dermis, but are rather straight close to the epidermis [6, 14–16]. Dick [16] observed that the elastin fibers in the main dermis layer are coarser than those close to the epidermis.

Reticulin fibers account for 0.4 percent of the dry weight of the skin [17]. They are located around blood vessels and hair, close to the epidermis [9]. Their chemical composition is similar to that of collagen [18]. Because of their minute quantity, they have little effect on the skin's overall response.

In response to uniaxial stretch the skin contracts in the lateral direction and the collagen fibers gradually align themselves in the direction of stretch. Their packing becomes denser and they subsequently stretch [6, 19, 20]. The fibers' alignment increases the strength and stiffness in the direction of stretch—an important merit in a tissue which may be stretched in various directions.

Stretch also affects the surface pattern of the skin. When skin is uniaxially stretched, the surface grooves align in the direction of stretch while the surface roughness in this direction decreases [21].

11.2 THE MECHANICAL ROLE OF THE SKIN COMPONENTS

Collagen fibers are the major mechanical elements in the skin. When these fibers are stretched, their effects predominate over those of all the other components. They are strong (tensile strength of 1.5 to 3.5 × 10^2 MPa) and stiff (Young's modulus in the linear region is approximately 1 GPa), and they stretch reversibly up to 2 to 4 percent. In the native state they are viscoelastic [22, 71, 113]. Thermodynamic studies have revealed that collagen behaves like a crystalline rather than a rubberlike material [23].

Elastin is considerably less stiff than collagen but can be reversibly stretched to more than 100 percent [24]. Its elastic properties are related to configurational entropic changes as in rubber and rubberlike materials [25]. On the basis of mechanoenzymatic studies [26], mechano-histological observations [16, 27, 28], and comparative mechanical data [15], it is apparent that the elastin fibers are the first to be stretched when the tissue is strained. Their effect on the response of the whole skin is thus significant at low levels of strain when the collagen fibers are still crimped. This notion is further strengthened by thermomechanical data [29] which show that at low strain levels skin has a negative thermal expansion coefficient (as in elastin) while at higher strain levels it is positive (as in collagen).

The role of the ground substance matrix in the response of the skin to tension is not yet clear. Studies [155] on the effects of temperature and humidity on the spectrum of relaxation times in very low strain levels (0.6 percent) show similar responses of skin to that of gels consisting of hyaluronic acid and water. This suggests that the ground substance is responsible for the viscoelastic behavior in this low strain range. It was also suggested that the fluidlike ground substance exudates from the interfiber space while the fibers become reoriented and densely packed upon stretch. This "flow" of ground substance may be associated with the viscous part of the skin's behavior, primarily in the "toe" region (the nonlinear region of the stress-strain curve), in which the collagen fibers are not yet fully aligned and stretched.

There exists, however, other evidence which suggests a lesser significance of the matrix. Study of the in vitro response of rat skin in which the hyaluronic acid (regarded as the predominant mechanical component in the ground substance) was differentially digested [30] revealed little effect on the rate of creep as compared with untreated controls. Similarly, Vlasblom [31] found no effect of the same treatment (hyaluronidase) on the rate of creep in torsional tests on human skin in vivo. Daly [32] found that in the initial reorientation phase (in which the migration of collagen fibers is thought to squeeze the fluid away), the response of the skin is approximately elastic. The viscous effect is highest at the final linear region when the collagen fibers are closely packed and straight.

Compressive tests of the skin (indentometry and skin-fold compressibility) show significant effects of the ground substance. The speed of compression [33–35], the amount of expressible fluid [34], and the skin's permeability [36] all increase upon application of hyaluronidase. These effects are thought to be related to the water affinity of the matrix mucopolysaccharides.

The epidermis contributes little to the skin's resistance to stretch [37, 38], but its contribution to the frictional resistance of the skin is predominant [39–41]. Its effect on the compressive indentation response of the skin is not clear [35, 42]. More detailed review of the role of the epidermis is given in Sec. 11.5.8.

11.3 IN VIVO STRESSES AND STRAINS

One of the most important features of the skin is its state of tension. Skin, in vivo, is normally under tension. This tension (expressed as force per unit length) varies in the range of 0 to 20 N/m [1, 27], depending on site, direction, and body posture. Upon excision, the skin will retract by 5 to 30 percent, again depending on site, direction, and posture.

Directional effects in the skin tension have since long been reported [43] and investigated in detail [20, 44]. It was found that circular punctures produce elliptical holes (splits) in the skin. The trajectories of the splits' major axes (cleavage lines) are called *Langer's lines* (Fig. 11.1). They are of considerable importance in plastic surgery.

Langer's cleavage lines are in fact intimately related to the visible crease and wrinkle lines of the skin, except on the palms, soles, and some areas on the limbs [44]. Flint [45] concluded that some earlier contradictory claims (stating that Langer's lines do not follow the same pattern as the wrinkles) were based on grossly inaccurate reproduction of Langer's original drawings and their misinterpretation.

Data on directional variability of in vivo tension and deformation [46–48] show that the highest stresses and strains are in the direction of Langer's lines. In addition, retraction of linear wounds is higher when the wounds are cut across Langer's lines rather than along them [44, 48, 49].

The skin's mechanical anisotropy follows the same directional pattern as Langer's lines: Its extensibility is lowest in the direction of the lines, both in vivo [50] and in vitro [20, 49, 51]. The skin's stiffness was shown to be highest in the direction of Langer's lines [51], but detailed observations show this to be true only at the low strain levels. At higher strains the slope in the linear region of the stress-strain curve is similar in all directions [20, 47–49].

11.3.1 Structural Aspects of Langer's Lines

Langer noticed that the cleavage lines correspond to the direction of preferred disposition of collagen fibers. Later, from histological observation [44, 47], it was concluded that both the collagen and elastin fibers are preferentially oriented in the lines' direction. Similar conclusions were drawn from scanning electron microscopic studies [7, 52].

Apart from the preferential disposition along Langer's lines, it was also found that the elastin [45] and the collagen [47] fibers along these lines are more stretched than those across the lines.

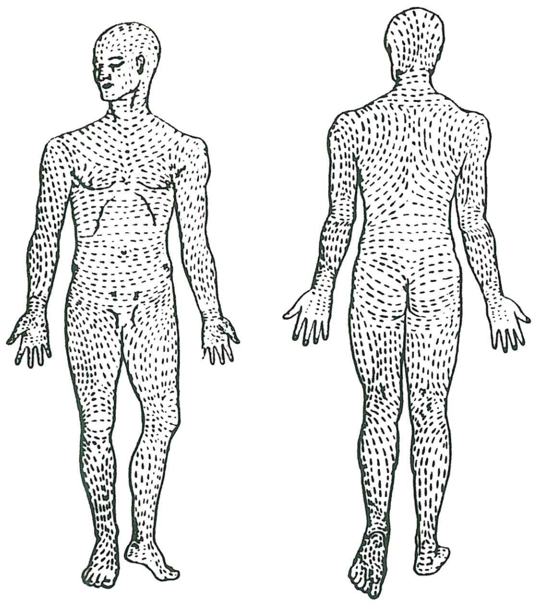

FIG. 11.1 Langer's cleavage lines. (*From Cox [44], with permission.*)

Consequently, the former are under higher tension and higher stretch, thus giving rise to a lower extensibility and to a higher initial stiffness in this direction—features which indeed characterize the skin in vivo.

It was suggested that the elastic fibers are the ones mainly responsible for the in vivo tension field. This hypothesis is supported by histological observation [16, 27, 45].

A recent study of the directional variability in the epidermis surface pattern [21] indicates that the major grooves (highest roughness) are along Langer's lines.

11.4 METHODS IN SKIN MECHANICS RESEARCH

Several engineering methods have been applied in the investigation of the mechanical properties of the skin, both in vivo and in vitro. An ideal procedure should consist of in vivo multiaxial tests with uniform strain field, such that the skin's comprehensive properties (the constitutive equations) can be readily associated with the results. This ideal has not yet been achieved. Under in vivo tests it is exceedingly difficult to obtain a uniform strain field throughout the specimen or to control the conditions along the boundaries. In addition, the resting tension and deformation [46, 53, 54], as well as the skin's thickness [55, 56], must be measured. Under in vitro conditions these experimental problems are easier to deal with. The disadvantage of in vitro tests is that the effects of blood and blood pressure, lymphatic drainage, in vivo metabolism, and nervous and hormonal controls are not present, and their significance cannot be evaluated. The most visible result of the altered environment in vitro is that the skin specimen swells [29].

Another problem common to many in vivo and in vitro tests is the attachment via grips. They may introduce local stress concentration or induce nonuniformity in the strain field. If slippage occurs, then the strain measurement based on the grips' displacement is inaccurate. The effect of the grips is particularly significant under in vivo tests in which stretching is performed via tabs which are glued to the epidermis [57]. Some of the preceding problems can be overcome by direct measurement of the tissue strain at sites sufficiently remote from the grips [29].

Although no overall superior experimental method is available, some techniques have distinguishing features which make them better suited for specific studies. Uniaxial in vivo tests [14, 57, 109, 114] are well suited for studies of directional effects, but the tissue's strain is not uniform. Improved strain uniformity can be achieved under either in vitro uniaxial tests [47, 51, 75, 98] or strip biaxial tests both in vivo [54] and in vitro [123]. In vivo suction tests with racetrack-shaped cups [53] and in vitro biaxial tests [29] are also associated with uniform strain fields and are well suited for studies of the anisotropic constitutive behavior of the skin. In vitro torsional tests [31, 126, 129, 130] and suction tests with circular cups, both in vivo [31] and in vitro [27], are easy to perform, but their interpretation in terms of the skin's properties is very difficult due to the highly nonuniform strain field. Skin-fold compressibility tests [153] and indentation tests [33, 39, 76, 100–102] are probably the easiest to perform in vivo. Although they cannot be used for investigation of the skin's constitutive behavior (owing to a highly nonuniform strain field), they have found extensive use in parametric studies of the compressive response of the skin, especially with regard to the effect of diseases and medicocosmetic agents.

Other details on experimental procedures can be found in previous reviews which have concentrated on in vivo techniques [58], on pharmacological aspects [59], and on hardware considerations [60].

Investigations of the epidermis and its stratum corneum were carried out with similar techniques to those used for the whole skin. Under in vivo conditions it is difficult to isolate the effects of the epidermis from those of the dermis. The stratum corneum can, however, be separated from the dermis and tested by itself. The separation can be done by a variety of methods which include heating to 60°C [61], suction blistering [62], chemical blistering with ammonia [63] or with cantharidin [64], and by enzymatic digestion [65]. It has been suggested [66], however, that some epidermal damage may be caused by these treatments. This topic requires further clarification.

11.5 GENERAL CHARACTERISTICS OF THE SKIN'S MECHANICAL RESPONSE

The functionally important mechanical properties of the skin are extensibility, resistance to friction, and response to lateral compressive loading. The first is associated mainly with the dermis and the second mainly with the epidermis, and the last is due to the combined effects of both.

Skin properties are highly variable, depending on species, age, exposure, hydration, obesity, disease, and biological difference between individuals. In the same individual, the skin's properties vary with site and orientation and may be altered by irradiation, drugs, and chemicals. Space limitations prevent the inclusion of an appropriate account of all the above effects in this chapter. Attention will be concentrated primarily on the general mechanical characteristics of the skin.

In the following, the important mechanical features of the dermis and epidermis will be outlined. A more quantitative outlook will be provided in the review of skin models.

11.5.1 In Vivo versus in Vitro Response

It is difficult to compare in vivo and in vitro responses of the same specimen both because of the poor control of boundary conditions in vivo and because of the physiological differences between in vivo and in vitro conditions. Reported results of such comparisons do not always agree: Cook et al. [53] found that in vivo tension is between approximately 10 percent (at low strain levels) and 20 percent (at high strain levels) higher than in vitro tension for the same strain in rat skin. Kenedi et al. [67] observed slightly higher loads in detached human skin specimens than in in situ specimens, but the differences were within their experimental error.

Vogel and Denkel [57] and Vogel [156] show that if the same gripping method is used both in vivo and in vitro, then the stress-strain relations are similar, with a slight difference in the strain-dependent anisotropic response.

11.5.2 Equilibrium Configuration and Preconditioning

Because of the skin's considerable slackness at low strain levels, it is difficult to establish a stress-free configuration for skin. If the specimen floats on a physiological solution, then the stress-free configuration can be established [29]. It depends, however, on the specimen's strain history. It has been recognized for some time that skin behaves in an approximately repetitive mode only after several test cycles. The stress in the second cycle is lower than in the first one, then it drops somewhat less in the third cycle, and so on until eventually a stable response is obtained. This is called *preconditioning*. It is accompanied by a parallel increase in the gage length in uniaxial tests, and by configurational changes in biaxial tests. In the case of biaxial stretching it was found [29] that each biaxial test requires its own preconditioning and is associated with its own equilibrium configuration. The effects of preconditioning are reversible if no damage has been caused. Recovery of both initial configuration and initial response will occur provided the specimen has not contracted in any planar direction [29].

The time constant for recovery from preconditioning is measured in hours. This is orders of magnitude higher than the characteristic viscoelastic relaxation time. It is thus possible to separate the two effects. This should, however, be done with caution, since skin may exhibit a wide spectrum of relaxation times.

Preconditioning affects the in vivo response as well. This was observed in uniaxial [68] and torsional tests [31, 69]. Preconditioning exists in other tissues [70], as well as in isolated collagen fibers [71].

From the modeling point of view, the effect of preconditioning must be included in any general constitutive formulation. This has not been done so far. Moreover, there seems to be no relevant data on this topic.

11.5.3 Homogeneity

Skin is obviously not a homogeneous material because of the fibrous, cellular, vascular, granular, and amorphous components of which it consists.

In practical problems we are usually interested in dimensions of specimens which are orders of magnitude larger than any of the skin's components. In many clinical problems, the response of the entire skin thickness is of importance. In these cases we may consider the skin as statistically homogeneous.

11.5.4 Compressibility

The question of whether skin can be considered incompressible is important since the constitutive formulation of incompressible materials is significantly simpler than in the general case. The evidence is, however, conflicting. Upon direct compressive loading [34], the skin will lose more than half its water at ~0.3 MPa. If however, hydrostatic pressure is used [72], then the compressibility (expressed by the bulk modulus) is 0.33 GPa, close to that of water (~0.25 GPa). Vossoughi and Vaishnav [73] point out that it is the relative compressibility (ratio of bulk to shear modulus) which determines whether a material can be considered incompressible. They showed this ratio in the skin to be of order 10^3, thus confirming its incompressibility.

Stretching tests in vitro show different results. Several investigators observed fluid secretion from specimens stretched in air [14, 74, 75].

The relevance of these in vitro observations to the in vivo state is not clear. Direct evidence of in vivo compressibility can be found in impedance tests [76]. The ratio of the bulk to the shear modulus was found to be of order 10^6. Although this points to in vivo incompressibility, more evidence is needed before conclusions can be drawn.

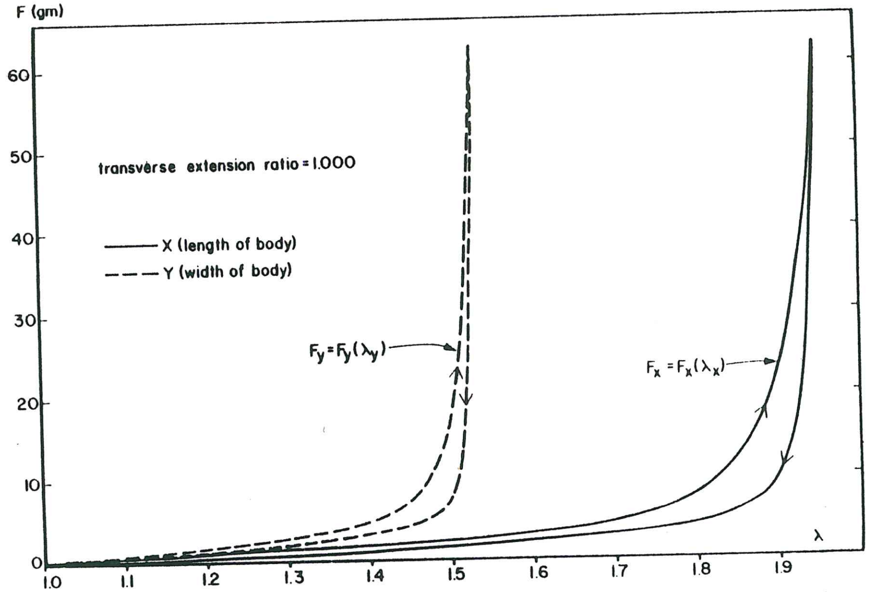

FIG. 11.2 Load-strain relations of a square (3.5 × 3.5 cm) specimen of rabbit abdominal skin. The tissue is stretched in one direction while the transverse dimension is kept constant. Solid line: stretch along the body length. Dashed line: stretch along the body width. (*From Lanir and Fung, [29], with permission.*)

11.5.5 Stress-Strain Relations

The stress-strain relations of the skin are both nonlinear and anisotropic (Fig. 11.2). Initially the skin is slack, and small stresses produce large strains. With further stretch the stiffness increases up to a maximum constant level in the linear region. Further stress will result in failure. The skin's anisotropy is evident from the different responses in different directions, both in vitro and in vivo, and under uniaxial as well as biaxial stretching.

Under biaxial homogeneous stretch along and across Langer's lines, both in vitro [29] and in vivo [46], no shear strains were observed. This points to some symmetry in the skin's response, namely, orthotropy or transverse isotropy.

As mentioned earlier, the nonlinearity of the skin's response can be readily accounted for by the rotation and gradual straightening of the collagen fibers. In the high-strain region they are all straight, and the stress-strain curve becomes linear. Under uniaxial stretch it is often observed [20] that the linear slope is equal in all directions. This is probably associated with a complete alignment of the fibers in the direction of stretch.

11.5.6 Viscoelasticity

Skin is a viscoelastic tissue. Its stress-strain relations are rate-dependent and exhibit considerable hysteresis. Skin tissue shows stress relaxation under constant strain and creep under constant stress. The skin's viscoelastic nature has been demonstrated in uniaxial, biaxial, torsional, and various dynamic compressive tests (penetration, elevation, skin-fold compression, ballistometry, and acoustic impedance).

The viscoelastic response of the skin is not linear: the modes of relaxation and creep depend on the corresponding levels of strain or stress in both uniaxial [15, 77, 78] and biaxial testing

(Fig. 11.3*a*). At low strain levels the skin may behave in an approximately elastic manner both in vitro [26] and in vivo [54]. Under in vitro creep testing, it was shown that following a transient time of several minutes the extension rate becomes constant and is proportional to the stress [79].

In vitro biaxial stress-relaxation tests reveal [29, 80] that two principal stress components relax in different modes (Fig. 11.3*b*). The skin is thus anisotropically viscoelastic. The above tests also reveal that under uniaxial stress-relaxation tests, the lateral dimension continuously contracts following the application of stress.

11.5.7 Damage and Strength

Although total rupture of healthy skin is not common, increased tension may cause various degrees of damage. Ruptures due to penetration of sharp objects occur in animals [2] and in humans during accidents. The skin's penetration failure under moderately sharp edges was found to be of a tensile (rather than shear) character and independent of edge shape and rate of loading [81].

Partial damage to the skin is more common, as demonstrated by the striae. *Striae* are stretch lines that occur under moderate tensions (not high enough to cause arrest of blood supply—blanching) which persist for a long time. They are common in pregnancy and may also occur over the muscles in excessive body-building exercises and in rapidly enlarging breasts [82]. At higher tension levels, blanching of the skin and subsequent necrosis will occur. Some skin damage in the form of permanent stretch may occur under tensions lower than those which cause striae.

The tensile strength of skin has typical values in the range of 2.5 to 16 MPa [83]. It varies with site [84], is higher in the main fiber direction [85–87], is higher in males than in females [85], and is believed to increase with age [83, 85, 88, 89], but contrary trends have also been reported [90]. Skin may also break under considerably lower stresses (as low as one-fifth of the tensile strength) under long creep tests [79, 87, 89].

11.5.8 Mechanical Properties of the Epidermis

The epidermis is exposed to a wide range of temperature and humidity, whose effects on its mechanical response are of prime interest.

Epidermal Stress-Strain Relations. The stress-strain relations are in fact significantly influenced by both humidity and temperature [91]. At low humidities (25 to 30% relative humidity), the curve is linear up to rupture, with a slope (Young's modulus) of order 1 to 10 GPa. But at 100% relative humidity the curve has between two [92] and three [93] linear regions with an initial slope which is 1000-fold smaller than in the dry state [92, 94]. A similar drop of the slope is caused by higher temperatures [92, 95]. The stress-strain data have a considerable scatter which can be reduced by a suitable moistening regime [94]. Under in vivo conditions the epidermis is hydrated only on its dermal side. Its properties are then different than in the fully hydrated state [96].

Epidermal Strength. The tensile strength was found to decrease, and the breaking strain to increase, with increasing humidity and temperature [91–93]. The strength is of order 10^2 MPa at low (26%) humidities and 10 MPa in the wet state. There is a similar drop with temperature in the 25 to 60°C range [92]. The breaking strain increases with humidity by close to two orders of magnitude, from approximately 2 percent at 26% relative humidity to 150 to 200 percent if wet [92].

The overall contribution of the thin stratum corneum is small compared with that of the thick dermis [37, 38]. In rupture tests the stratum corneum was observed to tear before the dermis [83].

Epidermal Viscoelasticity. The stratum corneum is viscoelastic. Its stress-strain relations are rate-dependent, becoming stiffer with increasing strain rate [4]. In dynamic in vitro tests the

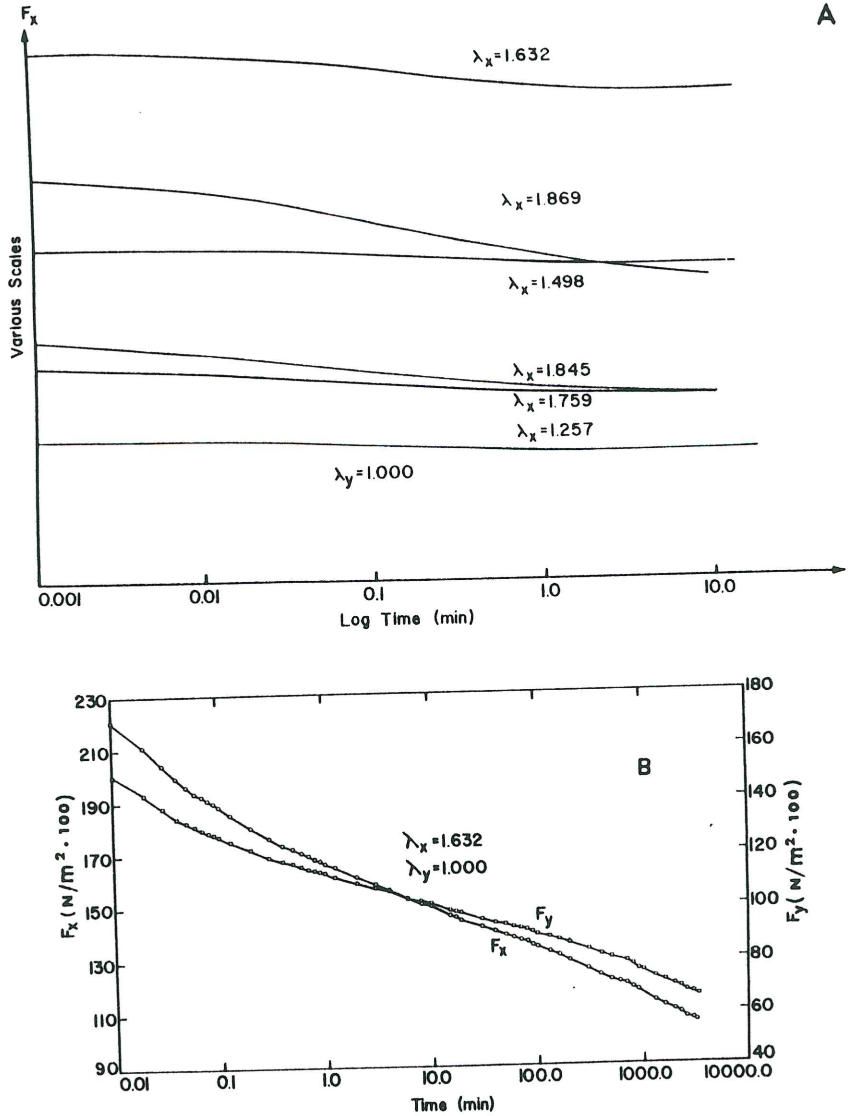

FIG. 11.3 (*A*) Mode of stress-relaxation in a square (3.5 × 3.5 cm) specimen of rabbit abdominal skin following a step stretch in the *x* (body length) direction, at various stretch ratios (λ_x). The transverse dimension is kept constant (λ_y = 1.000). (*B*) A comparison between the simultaneous relaxations of two principal stress components following a step stretch. The stretch ratios are indicated. (*From Lanir and Fung [29], with permission.*)

isolated stratum corneum shows storage and loss properties. Both depend on strain, frequency, humidity, and temperature. The loss (E'') and damping (tan δ) moduli were found to be highest in the 50 to 80% relative humidity range in human specimens [97], while the storage modulus (E') drops continuously with increasing humidity. Temperature has a dramatic effect on the storage ai lamping moduli in the physiological range but much less effect on the loss modulus. There ai lamping peaks at about 10 and 40°C. Frequency has only a small effect. Both E' and E'' in ase slightly with frequency, while $\tan\delta$ is relatively unaffected in the 3.5 to 110-Hz range [9 The dynamic modulus E' of human stratum corneum under 72% relative humidity, at 30°C, ai at 200 Hz was found [157] to increase with strain initially, peak at a strain level of ~0.20, ai drop gradually thereafter. E' is unaffected by strain rate and by preconditioning cycles and rc ains constant throughout the stress-relaxation process following a step change in the static st n.

F ctional Properties. The frictional properties of the epidermis are of considerable interest in n ny fields (e.g., the textile and shaving industries). Unfortunately, the available results are fc and at times conflicting. Well-controlled in vivo tests [41] revealed linear relations between fi tional force and lateral load (Amonton's law) and no difference between dynamic and static c fficients of friction (0.5 with polyethylene). In vitro tests, however, showed a lower dynamic c efficient and a reduction of the coefficient of friction with load for several textile materials [ ].

The frictional properties of the epidermis depend on several environmental factors. The c efficient was shown to increase by sweat or moderate moistening and by the removal of g ease, but was reduced by wetting, by dry talc powder, and by addition of oil [41]. The effect of v tting in vivo varies with time [39].

11.5.9 The Skin's Response to Lateral Loading

T e response of the skin to lateral compressive loading is of considerable functional and clinical in portance. Studies of this subject are based primarily on indentation and skin-fold compress-ib ity tests. Elevation tests have been used as well [121], but they are associated primarily with st etching deformation and consequently yield different results [100]. Most of the compressive st dies on the skin were done in vivo. The main results are discussed in the following subsec-ti ns.

E fects of Dermis and Epidermis on Compressive Response. The skin's response to compres-si e loading involves both the dermis and the epidermis. In some previous reports it was o served that certain cosmetic agents (including moisturizers) modified the tissue's response u ler specific procedures. It was thus claimed that these tests are associated primarily with the e dermis. Tests which involve the removal of the stratum corneum yielded conflicting results. In some [42] the denuded skin had a different response from the intact specimen, while in a r ent study [35] no difference was observed. It can be seen that the roles of the dermis and e dermis under compressive loadings are not yet clear and this area of study requires further i estigation.

C mpressive Viscoelastic Effects. The compressive response of the skin is highly viscoelastic [, 101, 102]. It is also nonlinear: the stiffness was found to increase with indentation [100]. I th the viscous and the nonlinear aspects of the skin response were found to diminish consid- bly by preload [103]. The immediate (elastic) and delayed (viscous) responses of the skin d end on age, sex, site, hydration, and obesity. A treatment with hyaluronidase was found to in rease both the magnitude [35] and the speed [33] of indentation. The magnitude also in- c ased after treatment with elastase but decreased after injection of saline. It did not change llowing treatment with collagenase [35]. These results indicate that the tighter elastin fibers d the ground substance may be the predominant components in the response to compressive ading and that collagen has little or no effect.

C mpressive Dynamic Response. The dynamic response of the skin is related to its capacity to sorb impact energy in accidents. Most of the previous studies used the normal vibrational pedance characteristics of the skin as the key to its dynamic behavior. Theory and experi- nts show [76, 104] that shear waves predominate at frequencies up to 5 kHz while compres- e waves are more important thereafter. Surface waves which can be observed on the skin are ombination of both.

The impedance characteristics were found to depend on frequency, humidity, probe area, and site on the body, as well as on the static pressure [104]. If the shear impedance is expressed as a complex number ($\mu = \mu_1 + i\mu_2 w$, where w is frequency and $i = \sqrt{-1}$), then typical values were $\mu_1 = 2.5 \times 10^3$ Pa, $\mu_2 = 15$ Pa · s [76]. In another study [158] the corresponding values at 600 to 1000 Hz were 1.5×10^3 Pa and 5.0 Pa · s. The impedance and phase angle were found [105] to increase linearly with the logarithm of frequency, but the precise relations depend on the preload. Increased humidity was found [159] to increase significantly the absorbed energy of human skin in vivo in the frequency range of 700 to 800 Hz.

In Vitro Compressive Response. A compressive study at physiological pressure ranges [106] revealed linear stress-strain relations up to a pressure of 7.8×10^3 Pa and an exponential relation (measured up to a pressure of 1.75×10^5 Pa). The behavior under in vitro cyclic compressive loading was found in the same study to be linearly viscoelastic at low pressures. At higher pressures the viscoelastic parameters decayed continuously and linearly with the logarithm of time (or cycle number).

11.5.10 Binding Forces in the Skin

The binding forces between different layers of the skin are of significance in its in vivo response, but little is known about this topic. The binding of dermis to subcutaneous tissues varies in different sites on the body (e.g., auditory canal versus penis). Cook et al. [53] found that the binding of dermis to subcutaneous tissues in the torso of the rat affects the results of suction tests by 10 to 20 percent. This was concluded from a comparison of in vivo and in vitro tests of the same specimen. Vlasblom [31] observed that a sliding load of 8×10^{-3} N was required to displace 1.0 cm² of human forearm skin by 1.0 mm. This introduced only a small (< 10 percent) error in his in vivo torsion tests.

The dermal-epidermal adhesion was found to be of a predominantly viscous nature [107]. The product of suction pressure and time to blister was found in this work to be approximately constant and equal to 90 ± 30% MPa · s in the lower abdominal human skin.

The binding forces between layers of the stratum corneum vary with site and sex. They are of order 2×10^4 Pa [108].

11.6 MODELS OF SKIN MECHANICS

Three major classes of models have been developed for the skin. In the continuum models, general material theories were specialized for the skin and used to simulate its multiaxial response. The phenomenological models consist of mathematical formulas describing the skin's behavior under specific deformational schemes. Some of them consist of mechanical elements such as springs and dashpots. In the structural models the skin's behavior is analyzed in terms of the combined effects of its components.

General multiaxial theories for viscoelastic behavior are available. They are, however, much too complex for practical use. In some cases, simpler approaches are adequate. One simplification widely used in skin research is based on the fact that under constant rate of stretch the stress-strain relations, in both the extension (ascending) and contraction (descending) phases, are unique for a preconditioned specimen. The skin can thus be regarded as pseudoelastic in these widely used tests.

The pseudoelastic stress-strain relations can be expressed as follows:

$$S^{ij} = \frac{\partial W}{\partial \varepsilon_{ij}} \qquad i, j = 1, 2, 3 \tag{11.1}$$

where $W(\varepsilon_{ij})$ is a function of the strain components and is known as the *strain energy function* (per unit original volume). S^{ij} and ε_{ij} are Kirchhoff's stress and Green's strain components, respectively. They are referred to the original dimensions of the specimen. W can also be expressed in terms of the three principal stretch ratios $-\lambda_i$.