Evaluation of Biomechanical Properties of Human Skin

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With the exception of the skeleton and the musculature, the skin is the largest organ, being about 1.7 m² in area and approximately 4 kg in weight, or about 5.5% body mass. Because it represents the outward appearance, skin has great social importance, but only when affected by serious disease do we fully realize the extent and importance of its many other, more basic functions.

Our skin is multifunctional and has consequently an extremely complex structure. It is multilayered with convoluted and often indistinct interlayer boundaries, and its properties are different in different directions. Its structure and appropriate function vary with body site. The properties of skin also vary with the rate of application of stress and the length of time over which the stress is maintained, and are very sensitive to ambient conditions, age, and recent handling. In vivo, any particular area of skin has a complex attachment to underlying tissues and to all the surrounding skin and contains and is crossed by its adnexae, epidermal appendages, and vascular and neural tissues.

The skin has important protective functions against mechanical trauma such as friction, impact, pressure, cutting, and shear. It must be an active (ie, adaptable) barrier between physiologic conditions within the body and the varying, unfriendly ambient environment. While also helping to regulate that internal environment by, for example, dissipating or conserving heat. Internal body structures must have a controlled freedom of movement within, but also have some support from, the skin which must act as the nonslip intermediate surface when we grip, lift, or press (eg, walking). All these functions are performed simultaneously by a physical structure that also has important physiologic and immunologic roles and that renews itself every 2 to 3 weeks.

Disease or dysfunction may affect some or all of the physical properties in one way or another, which may in some instances be the cause of some of the symptoms associated with a particular disease. Other causes of variation in physical properties include maturation and aging, exposure to solar radiation, and chemical, physical, and xenobiotic insult.

What, then, are the important physical properties of skin that are amenable to measurement? The stretch properties, tensile properties, are of great interest, as they are experienced in vivo. Breaking strength, elasticity, and the deformation and flow properties are those familiar to engineering materials testing, so industrial tensile tests, or adaptations of these, are often applied to skin measurement.

Measurements of Biomechanical Properties

Here we must consider techniques and interpretation of results in the two distinct categories of in vitro and in vivo test methodologies.

Results from in vitro tests are estimates of properties that are assumed to have an important influence on function in vivo, and can inform about the basic mechanical properties of strength, elasticity, or density inherent in the architecture of skin. In vivo tests tell us how the skin reacts to external forces of stretch, shear, torsion, compression, and indentation when it is on the body. In vitro tests can be considered as "once only" tests in that the specimen, once removed, cannot be further modulated in vivo. The serial testing of one site possible with nondestructive in vivo tests may provide information on the function and the kinetics of change in mechanical properties. The numerical values given by in vitro tests usually exhibit a smaller spread than those from in vivo tests, but in both cases objective data are available to support (or supplant) subjective assessments of the physical state of the skin. In vitro tests, unfortunately, test properties rather than function.

In Vitro Methods

In vitro methods involve the removal of skin samples from the body, which are tested as is any other material. The usual procedure is to excise the skin and to pare off as much subcutaneous fat as possible. The site and orientation of the specimen are very important, as anisotropic preexisting or "testing" tension of skin exists as a result of structural strain, normal habitual body movements, and underlying joints or muscular-
ture. Such resting tension gives rise to the characteristic gaping of wounds, where a linear cut across the lines of tension produces an elliptical wound. The lines of maximum tension were originally mapped by and are named for Langer in the late 19th century. On excision care must be exercised to minimize stretching of the sample, as skin exhibits a “memory,” so that the effects of strain are still obvious many hours after removal of the stretching force. The sample is cut or stamped into a dumbbell shape, the middle portion of which is considered the sample under test (the gauge) and the larger ends are gripped between specially adapted jaws of a tensile testing machine (eg, Instron). In this arrangement, when the specimen is stretched, the strain is uniform throughout the skin thickness, an important consideration when comparing in vitro with in vivo measurements.

There are two basic ways to measure tensile properties. The first involves applying all the strain at once, with minimum time between the steady states of zero load and maximum load. This method is useful to study stress relaxation (the way the specimen alters its structure, dimensions, or properties to minimize the stress), but the “instant” application of load is difficult. Also, skin exhibits a rate-dependent resistance to applied stress, and if load is applied too fast it may in fact rupture at stress levels much lower than the ultimate stress levels determined by other tests. The other method involves the application of load in very small increments, with time delays between successive load increases long enough for all stress relaxation to have ceased, or by using a constant but slow increase in load, a typical strain rate being 5 cm/min. This gives more consistent and reproducible results for the linear elastic portion of the skin response.

By careful control of the specimen temperature and humidity, and the extension regimen, and by good dimensional measurements, reliable and reproducible results can be obtained.

Information about the mechanical properties of skin available from in vitro testing falls under three broad categories: strength values (eg, breaking strain); time-dependent values (eg, creep, relaxation); non-time-dependent values (eg, elasticity).

When skin is stretch in a slow manner as described above, a graph of stretch force per unit cross-sectional area (stress) against applied extension per original gauge length (strain) can be constructed. This stress-strain curve exhibits an initial short portion, where elongation occurs without appreciable force. After this the curve increases in a roughly exponential manner until, at high extensions, it becomes almost a straight line indicating Hookean (purely elastic) behavior immediately before rupture occurs. If the initial part of this curve is magnified, a straight-line approximation will allow the calculation of elastic properties such as Young’s modulus. This low extension part of the in vitro stress–strain curve corresponds to the sort of extensions possible in in vivo experiments, and in experiments where both in vitro and in vivo tests have been done on the same (animal) subjects, the two agree fairly well.

The following section summarizes published results of careful in vitro testing of normal human skin samples of dimensions 4 mm long × 2 mm wide × skin thickness (approximately 1–2 mm), and relies heavily on the work of Vogel.

The tensile strength of skin (ultimate load divided by cross-sectional area) ranges from 5 to 30 N/mm², with the mean showing a maximum of about 21 N/mm² at 8 years, declining to about 17 N/mm² at 95 years.

The ultimate modulus of elasticity (calculated from the end portion of the stress–strain curves) ranges from about 15 to about 150 N/mm². The mean shows a maximum value of about 70 N/mm² at age 11, with a shallow decline to about 60 N/mm² at 95 years.

The ultimate strain (extension before rupture) varies from about 35 to 115%. The mean value declines in a linear fashion from 75% to birth to 60% at 90 years.

When the skin is stretched it immediately begins to adapt its internal collagen meshwork structure to minimize the strain. Thus, the load required to maintain a given extension gradually reduces to a minimum value; this is termed relaxation. Conversely, for a given load the specimen will continue extending after application, a phenomenon called creep. If skin samples are stretched at a steady rate then unloaded with the same velocity, it is found that the stress–strain curves for the extension do not coincide with the curves for the unloading cycle, forming what are termed hysteresis loops. The area under the stress–strain curves represents the energy put into the system, and the area between the loading and unloading curves represents the energy used or lost in the system. The relative amount of energy loss is a measure of the viscosity of the sample (a “fat” hysteresis loop represents a highly viscous material, while coincidence of the loading and unloading curves represents purely elastic behavior such as would be seen using a rubber band). In general, this viscosity parameter decreases from youth to senescence.

The extension still present after each unloading cycle (the residual tension) indicates the plasticity of the skin, and is not age dependent; however, other parameters indicating elasticity and strength show increases during maturation and a plateau (for low extensions) or a reduction (for high extensions) during aging.

In experiments where the skin is extended at a very high rate and maintained at the new extension, the relaxation phenomenon causes the stress values to drop with a roughly logarithmic rate after initial strain.
application. This loss of tension (measured a set time after application of the initial extension) expressed as a percentage of maximum initial tension indicates the plasticity of skin. Using this parameter skin shows a clear loss of plasticity with age. Other parameters of plastic or viscous behavior in skin show a pattern similar to those found in the hysteresis experiments, where the extension rates are slower.

If a skin sample is loaded with a stretching force and the resultant length increase measured over time, it shows an initial sudden extension followed by a slow and more or less continuous extension (creep). For example, using a 200-g load, this final extension rate is about 0.03% per minute. Using a 500-g load, the average rate declines from about 0.06% per minute for younger skin to about 0.03% per minute for skin of 90 years. The extension after 1 hour, which includes the initial extension, varies from about 36% in the youngest specimens to about 30% for the oldest with 200-g load, and about 63 and 48%, respectively, for the 500-g load. This again shows the loss of plasticity (viscosity) with age.

Broad analysis of skin samples for soluble and insoluble (in 0.15 M and 0.5 M NaCl, and in citrate buffer of pH 3.7) collagens, glycosaminoglycans, and elastin show a correlation between tensile strength and insoluble and total collagen. Collagen content of skin is known to decrease with age. Viscosity parameters correlate with glycosaminoglycans and/or soluble collagen content in skin, but elastin content has not been found to correlate with any measured mechanical property of normal dermis or whole skin.

From this and other in vitro studies, it can be clearly stated that skin shows extremely complicated viscoelastic behavior. In general, however, a summary of properties and their age variation can be given: young skin is less protective against large strains than older skin; young skin is more viscous, or plastic, than older skin; older skin has a proportionately greater elastic region in its stress–strain behavior than younger skin. It also seems reasonable to conclude that the strength and elastic properties of skin are determined by the collagen content (specifically the "insoluble collagen" content), and not the elastin or ground substance components. Viscoelastic properties do seem to correlate with the soluble collagen and glycosaminoglycan content. As these substances probably contribute to the water binding capacity of skin, and as viscoelastic properties arise largely from the fluid (blood, lymph) and water content of skin, then the observed correlation is not unexpected.

As well as the advantages of being able to isolate and exactly control the skin or skin layer under test, the in vitro method does have some disadvantages. It is a "once only" method, which requires the radical removal of the skin, which cannot therefore be further modulated in vivo. Thus, time series measurements must use different sites, which means more samples are required for adequate interpretation of results. Such results are not readily translated to the in vivo state because of the complex interactions of surrounding tissues, blood flow, and restriction on the extent and rate of stress application in vivo.

In Vivo Methods

Because only the top boundary of skin is accessible to an in vivo test method, there is no chance of designing a "pure" or direct measurement of any single property of the dermal fibrous connective tissue. Therefore, investigators have devised many tests that apply such stresses as would normally be experienced, and measure some force, distance, or time parameter (or some combination of these). In this way empirical data can be gathered, and normal ranges established against which any single result can be compared.

In vivo tests can be classified as static, where a single modulating stimulus is applied and some resulting change is measured, or dynamic, where a cyclical stimulus is applied to the skin and the initial adapting and final steady-state reactions are monitored.

Static Tests

The simplest method is that of linear uniaxial extension. Here, tabs are attached to the skin with cyanoacrylate glue (or less satisfactorily with doublesided adhesive tape) and driven apart to give a preset extension (typically 30% extension). The force required to stretch the skin and maintain the new tab separation is recorded on a chart recorder. The separation results in an initial peak force, \( F_{pk} \), which immediately drops due to the adaptation of the collagen meshwork of the dermis (relaxation). The amount by which the force drops is termed the time-dependent or viscoelastic force component, \( F_{ve} \), and the level at which the force no longer changes is the non-time-dependent or elastic force, \( F_e \). The rate of force reduction can be derived from the force–time curve by modeling it as a simple logarithmic or exponential function, in which the time constant, \( \tau (\tau_s) \), is the time taken for the force curve to drop to \( 1/e \) (37%) of its initial value. The values \( \tau \), \( F_{ve} \), and \( F_e \) describe the force–time curve to a first approximation, and may be related to skin behavior, and changes in one or more of these are characteristic of, for example, solar elastotic degenerative change, congenital disorders of connective tissues (eg, scleroderma, Ehlers–Danlos syndrome, cutis laxa), and orientation of the test in normal skin (this reflects the preexisting or "resting" tension in skin). An alternative method applies a constant extension force and then monitors the separation of the tabs. Modulation of stratum corneum properties (by as much as 30%), such as results from the effects of emollient applica-
sclerodermatous from normal skin on the forehead. A device called a durometer, designed to measure hardness, was used to determine some basic elastic and viscoelastic properties of skin and to assess skin hardness in scleroderma. A closely related method attempts to supply a biaxial or two-dimensional strain to the skin. This cannot easily be achieved in vivo, but a good approximation is obtained by using attachment tabs that are much wider than their initial separation distance. When they are stuck to the skin and moved apart, the skin is prevented from "necking" in the region of strain application and, therefore, experiences an effective stress in a direction orthogonal to the extension axis. This technique has been used in the study of resting tension in skin in vivo.

The response of skin to shear forces can be measured by applying a torsional stimulus. A typical twisting device has a central disk within a thin annulus. These are stuck to the skin and the inner disk of skin is twisted through a few degrees (usually 10° or less) while the torque required to achieve and maintain this rotation is monitored (or a set torque is applied and the angular displacement monitored). In the latest designs, a third annulus surrounds these elements and is used in an attempt to isolate the area under test from the surrounding skin. The creep and relaxational behavior of skin in this test is much the same as when using an uniaxial extensometer. The difference between the two methods lies in the fact that the torsional device twists skin through a small angle in all orientations on the skin, thus eliminating differences due to test direction. The torsional method has been used to determine some basic elastic and viscoelastic properties of skin and in studies of chronic sun exposure, racial differences in skin properties, scleroderma, and effects of cosmetic product application.

Another group of techniques designed to be direction insensitive are the indentation methods first described in 1912. The deformation resulting from a preset indenting force can be recorded (indentometry), or the elevation caused by a preset pulling force is measured (leverometry). The methods have been much used in the study of aging and the elevation caused by a preset pulling force is measured (leverometry). The methods have been much used in the study of aging. An indentometer device called a durometer, designed to measure hardness of rubbers and other soft materials, has been used to assess skin thickness in scleroderma. Here the durometer reading was found to increase reliably in patients who were clinically judged to show greater induration; however, the device failed to distinguish sclerodermatous from normal skin on the forehead. A skin pulling device has been used to quantify changes in the deformability, extensibility, and elasticity of a range of connective tissue disorders such as cutis laxa, corticosteroid atrophy, acroscleroderma, polyfibromatosis, and scleromyxedema.

The latter methodology has resulted in the development of suction devices, which involve placing a suction cup or cylinder on the skin surface and using a negative pressure to raise a dome of skin. The pressure and the height of the dome are used to calculate elasticity parameters and have been shown to be age dependent. Such devices usually include the ability to alternate the pressure in a dynamic test mode.

Dynamic Tests
Repeating the application of test stresses (cycling) results in accommodation of the skin, and by cycling the applied stimulus a "steady-state" response to the test can be achieved, leading to good reproducibility of results. Such dynamic test methods also have the potential to provide more information than static tests; however, there is a potentially unlimited combination of frequencies, magnitude, attachment area, and other test stimulus parameters that may not be standardized or whose effects on the skin response may not be fully understood. This may result in difficulties in interpretation of data and certainly leads to the inability to compare results between different devices. Several methodologies for dynamic testing have been developed.

In the ballistometric test system the rebound pattern of a small hammer impacting onto skin is recorded and analyzed. Inevitably the recorded responses contain contributions from an undetermined volume of skin, plus the underlying structures of fat, muscle, or bone, but such regional data may be useful. Elastic wave propagation systems have in the main concentrated on the measurements of shear wave velocities and the rate at which they are dissipated during their passage in the skin. The application of the mechanical stimulus producing these waves is via a vibrating stylus or by a piezoelectric element. Detection of the resultant wave is also achieved by a separate stylus or element. This method represents a uniaxial dynamic test method, the axis being defined by the line between transmitter and detector of the shear waves. Shear wave propagation is effected by the sideways movement of one layer of the medium being transmitted via viscous coupling to the next layer. This means that those skin components most associated with the viscous properties of skin are most sensitively assessed using this technique. Water content of skin has a major effect on its viscous properties and a report using this technique has presented results that suggest a probable reduction in skin water content with age.

Dynamic suction methods apply a suction cup to the skin and generate a cyclical suction. The height to which the skin is raised on each cycle is recorded either...
from the position of the skin surface or in a more sophisticated design by using M-mode ultrasound to visualize the movement of different skin layers throughout the suction cycle. This method isolates a defined area of skin, and the ultrasound visualization allows individual layer responses to be observed. Of course, these layers are still connected to adjacent tissue, so their response is not unique to their inherent tissue properties. Other designs of dynamic suction devices have been used in the study of scleroderma, in the study of hydration of the skin in the context of plastic surgery procedures, and in the study of regional and age-related differences in women.

Mechanical impedance methods measure the resistance that the skin shows when forced into motion. This varies with the frequency of the applied force, the state of stress of the tissues, the mechanical properties of the tissues, and the mechanical coupling of the applicator to the skin. Sensitive instruments such as the electrodynamometer accurately measure the force on and displacement of a vibrating head that is attached to the skin. This arrangement results once again in measurement of an undetermined volume of skin, but regional differences in skin mechanical properties are obvious. Scleroderma has been studied using a resonance method where the skin surface is forced into a sinusoidal motion by a shaker device whose frequency is swept between 10 and 1000 Hz. The frequency and shape of the resonance peak are indicative of the elastic properties of the skin, and scleroderma was seen to cause a three-peak low-amplitude resonance pattern, suggesting more than one mechanical component was present in the affected areas.

Mechanical tests have been proposed for the objective assessment of severity and response to treatment in several diseases. For example, scleroderma has been quite extensively studied with a variety of mechanical testing techniques. Kalis et al have shown that the thickening and stiffening of skin in both generalized and localized scleroderma can be quantified. They found for progressive morpheic lesions, skin was about 60% thicker and 67% stiffer than unaffected areas. For generalized scleroderma, skin thickened by about 100%, and again was about 68% stiffer. By consideration of a simplified model for skin behavior, these authors concluded that the increase in stiffness in progressive scleroderma was probably a direct consequence of the increase in thickness, but in regressing lesions a change in the intrinsic elastic moduli was detected. Pierard and Lapierre found a reduction of up to 75% in skin extensibility in the vertical plane in acrosclerosis.

Conclusions

Measurement of the physical properties of skin may seem “esoteric” and of little relevance to clinical science; however, the noninvasive nature of available techniques provides unique opportunities for monitoring the effects of disease, drugs, or cosmetics over time on exactly the same area of skin.

In vitro testing gives repeatable standardized methods that can supply basic elastic and viscoelastic moduli for skin, which for low strain are comparable to results obtained from in vivo tests. Interpretation of in vivo tests can be difficult, as no analytical model has been developed that can relate measurements from these tests directly to basic skin properties. Each method and each implementation of that method has subtle differences from every other method so that results between studies are difficult to compare; however, the alternative to the bioengineering tests is the hand and eye, which provide subjective (and often biased), nonlinear, and notoriously variable data between individuals. Judging the severity of involvement, the changes due to treatment or progress of disease, and the efficacy of competing treatments is very difficult, even for an experienced clinician. Objective, numerical information on the effects of different active compounds and their formulations is essential if new preparations are to be optimized. Measurement of mechanical properties is one aspect of this important field of endeavor.

References