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Triphasic FE Modeling of the Skin Water Barrier

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Abstract. A skin–air model is developed to model the water barrier function of skin. The skin model is a porous solid saturated with a monovalent salt solution. The air model is a vapor diffusion model in non-moving air. *In vivo* measurements of water loss from human skin under varying ambient conditions are used to validate the model.

Key words: ion diffusion, matric potential, evaporation, stratum corneum, transepi-dermal water loss.

Nomenclature

$B^{\beta\gamma}$	frictional tensors.
c^f	molar concentration of fluid.
c^i	molar concentration of ionic component.
D^i	ion diffusion tensor.
D^v	vapor diffusion tensor.
E	Green–Lagrange strain tensor.
E	Young’s modulus.
F	deformation gradient tensor.
J	local volume change relative to the initial configuration.
K	hydraulic permeability tensor.
n^f	current fluid volume fraction.
n_0^f	initial fluid volume fraction.
n^i	current ion volume fraction.
p	skin pressure.
p^v	vapor pressure.
R	gas constant.
R^v	specific gas constant of vapor.
RH	relative humidity.
S	effective second Piola–Kirchoff stress.
T	absolute temperature.
TEWL	transepidermal water loss.
$\vec{v}^f$	fluid velocity.
$\vec{v}^i$	ionic velocity.
$\vec{v}^s$	solid velocity.
$\bar{V}^f$	molar volume of the fluid.

$\bar{V}^i$	molar volume of the ionized salt.
W_E	elastic energy density.
W_ψ	matric energy density.
W	energy density.
X	depth from surface of the skin in the initial configuration.
μ^β	chemical potential of constituent β per unit volume.
μ_0^β	reference chemical potential of constituent β .
$\bar{\mu}^\beta$	molar chemical potential of constituent β .
π	osmotic pressure.
σ	total Cauchy stress.
σ^s	partial Cauchy stress of the solid.
σ^f	partial Cauchy stress of the fluid.
σ^i	partial Cauchy stress of ionic constituent.
σ^e	effective Cauchy stress.
ψ	matric potential.

1. Introduction

Our body is a porous medium with a fluid volume percentage of well over 50%. In rest 5–20 g/m²h water is lost from the body to the environment: the transepidermal water loss (TEWL). This is about 20 times less than water evaporating from a free surface. Thus the human skin forms an effective barrier against excessive water loss, which is of vital importance to allow control of the internal aqueous balance of the body. This remarkable barrier function is carried out by a layer which is only 10–20 μ m thick: the stratum corneum.

To model how the stratum corneum governs the water loss from the body to the environment so effectively, we propose a porous medium finite element model which simulates fluid- and ion flow through the skin. We quantify the several skin material parameters on data from literature and own experimental data. The model is verified using our experimental data of transient behavior of human *in vivo* skin. Most models of skin restrict their field of application either to deformation or to transport phenomena. Oomens *et al.* (1987) are the first authors to incorporate transport of fluid and deformation of skin in a single sponge-like mixture model. Along the lines of the work of Lai *et al.* (1991) on articular cartilage and Snijders *et al.* (1995) on the inter-vertebral disk, we introduce a triphasic model of skin in contact with air to model water loss. The model includes a solid subject to finite deformation, water, and an ionic component coupled with a diffusional model for water vapor transport in air. Driving forces for transport are gradients in chemical potentials of the fluid and the ions. Electrical effects are not taken into account. Own measurement (unpublished data) of the fixed charge density of one sample of rat skin with cation tracer technique, showed a fixed charge density of 0.015 Meq. Pikal and Shah (1990) fitted a fixed charge density based on measurement of transport numbers and found 0.013 Meq. Both values are an order of magnitude lower than those measured in cartilagenous tissues, favoring our choice to neglect

electrical effects. The transepidermal water loss, is thought to be dictated by the difference in vapor pressure between skin surface and the environment. The vapor pressure just above the skin surface is related to the chemical potential of the water at the skin surface. The approach consists of a skin and an air model coupled by equating corresponding chemical potentials of water at the surface of the skin to vapor chemical potentials just above the skin surface.

2. Model

2.1. TRIPHASIC MIXTURE MODEL FOR SKIN

The triphasic mixture model for skin consists of three components: a solid (s), a fluid (f), and an ionic component (i). The behavior of the mixture is the result of the behavior of the distinct components and the interaction between them. Assuming saturation, all components intrinsically incompressible, excluding mass transfer between components, and neglecting the volume fraction of the ions relative to the other volume fractions, the volume balance of the mixture reads

$$\vec{\nabla} \cdot \vec{v}^s + \vec{\nabla} \cdot (n^f (\vec{v}^f - \vec{v}^s)) = 0, \quad (1)$$

where $\vec{v}^\alpha$ is the Eulerian velocity, and n^α the volume fraction of component α . The volume balance of the ions reads

$$\frac{\partial n^i}{\partial t} + \vec{\nabla} \cdot (n^i \vec{v}^i) = 0. \quad (2)$$

Neglecting body forces and inertia, the momentum balance of the mixture reads

$$\vec{\nabla} \cdot (\boldsymbol{\sigma}^s + \boldsymbol{\sigma}^f + \boldsymbol{\sigma}^i) = \vec{\nabla} \cdot \boldsymbol{\sigma} = \vec{0}, \quad (3)$$

where $\boldsymbol{\sigma}$ is the Cauchy stress tensor.

To arrive at a complete set of equations for the description of the behavior of a triphasic mixture, constitutive relations for the components and for the interaction between components are introduced. By means of the entropy inequality for isothermal and incompressible conditions, which holds for an arbitrary state of the mixture, complying with the balance laws, restrictions on the constitutive relations can be derived. The independent variables are the Green–Lagrange strain $\mathbf{E}$, the current volume fractions of the fluid per initial mixture volume Jn^f , the current volume fraction of the ions per initial mixture volume Jn^i , and the relative velocities $\vec{v}^f - \vec{v}^i$ and $\vec{v}^s - \vec{v}^i$ (Lai *et al.*, 1991; Snijders, 1994; Huyghe and Janssen, 1997). $J = \det \mathbf{F}$ represents the volume change of the mixture relative to the initial state.

The total Cauchy stress is written as

$$\boldsymbol{\sigma} = \boldsymbol{\sigma}^e - p\mathbf{I}, \quad (4)$$

where $\boldsymbol{\sigma}^e$ is the effective Cauchy stress tensor and p the pressure. Assuming elastic behavior, the effective Cauchy stress tensor is given by

$$\boldsymbol{\sigma}^e = \frac{1}{J} \mathbf{F} \cdot \frac{\partial W}{\partial \mathbf{E}} \cdot \mathbf{F}^c, \quad (5)$$

where $W = W(\mathbf{E}, Jn^f, Jn^i)$ is the strain energy function and the del operator signifies that the derivative is a partial derivative, that is, the fluid volume fraction Jn^f and the ion volume fraction Jn^i are kept constant. The chemical potentials of the fluid (μ^f) and the ions (μ^i) are, respectively,

$$\mu^f = \frac{\partial W}{\partial (Jn^f)} + p, \quad (6)$$

$$\mu^i = \frac{\partial W}{\partial (Jn^i)}. \quad (7)$$

The entropy inequality reveals that W depends on the Green–Lagrange strain tensor $\mathbf{E} = \frac{1}{2}(\mathbf{F}^c \cdot \mathbf{F} - \mathbf{I})$ and the volume fractions Jn^f and Jn^i only.

Transport of fluid and ions is caused by gradients in their chemical potentials. Using the entropy inequality the classical equation of irreversible thermodynamics is derived

$$-n^\beta \vec{\nabla} \mu^\beta = \sum_{\gamma=f,i} \mathbf{B}^{\beta\gamma} \cdot (\vec{v}^\gamma - \vec{v}^s), \quad \beta = f, i, \quad (8)$$

where $\mathbf{B}^{\beta\gamma}$ is a positive definite matrix of frictional tensors. Equation (8) describes the transport of fluid and ions. It consists of two equations

$$-n^f \vec{\nabla} \mu^f = \mathbf{B}^{ff} \cdot (\vec{v}^f - \vec{v}^s) + \mathbf{B}^{fi} \cdot (\vec{v}^i - \vec{v}^s), \quad (9)$$

$$-n^i \vec{\nabla} \mu^i = \mathbf{B}^{if} \cdot (\vec{v}^f - \vec{v}^s) + \mathbf{B}^{ii} \cdot (\vec{v}^i - \vec{v}^s). \quad (10)$$

Adding these equations and assuming $\mathbf{B}^{if} = -\mathbf{B}^{ii}$ (which physically means that friction between the ions and the solid is neglected), yields

$$-(\mathbf{B}^{ff} - \mathbf{B}^{ii}) \cdot (\vec{v}^f - \vec{v}^s) = n^f \vec{\nabla} \mu^f + n^i \vec{\nabla} \mu^i. \quad (11)$$

By introducing the permeability tensor

$$\mathbf{K} = (n^f)^2 (\mathbf{B}^{ff} - \mathbf{B}^{ii})^{-1}, \quad (12)$$

Equation (11) can be written as

$$n^f (\vec{v}^f - \vec{v}^s) = -\mathbf{K} \cdot \left(\vec{\nabla} \mu^f + \frac{n^i}{n^f} \vec{\nabla} \mu^i \right). \quad (13)$$

Assuming a strain energy function W of the form

$$W = W_E(\mathbf{E}) + W_\psi(Jn^f) + \frac{RT Jn^i}{\bar{V}^i} \left(\ln \left(\frac{n^i}{n^f \bar{V}^i} \right) - 1 \right), \quad (14)$$

where W_E is the elastic energy function and W_ψ the matric energy function, Equations (6) and (7) yield classical expressions for the chemical potentials

$$\mu^f = \mu_0^f + p + \psi + \pi = \mu_0^f + p + \psi + \frac{RT}{\bar{V}^f} \ln(c^f), \quad (15)$$

$$\mu^i = \mu_0^i + \frac{RT}{\bar{V}^i} \ln(c^i), \quad (16)$$

where μ_0^β is the chemical potential in a reference state, p the fluid pressure, ψ the matric potential ($\psi = \partial W_\psi / \partial (J n^f)$) (Nitao and Bear, 1996), π the osmotic pressure, R the universal gas constant, T the absolute temperature, $\bar{V}^\beta$ the partial molar volume, and c^β the concentration of component β per unit fluid volume.

The matric potential accounts for fluid–solid interaction (capillary and adsorptive effects) (Nitao and Bear, 1996). In many porous media, for example, articular cartilage and intervertebral disco tissue, the matric potential is fairly constant and is therefore often omitted. However, for skin in contact with air the matric potential plays an important role.

Substitution of Equation (10) into Equation (2) (assuming $\mathbf{B}^{if} = -\mathbf{B}^{ii}$) yields the diffusion equation for the ions.

$$\frac{\partial n^i}{\partial t} + \vec{\nabla} \cdot (n^i \vec{v}^f) = \vec{\nabla} \cdot \mathbf{D}^i \cdot \vec{\nabla} \mu^i, \quad (17)$$

where $\mathbf{D}^i = (n^i)^2 (\mathbf{B}^{ii})^{-1}$ is the diffusion tensor of the ions.

The equations describing the behavior of a triphasic mixture (Eqs. (3), (1), and (2)) combined with the constitutive relations (Equations (4), (13) and (10)) lead to a set of three coupled differential equations in which geometric and physical nonlinearities occur:

$$\vec{\nabla} \cdot \boldsymbol{\sigma}^e - \vec{\nabla} p = \vec{0}, \quad (18)$$

$$\vec{\nabla} \cdot \vec{v}^s - \vec{\nabla} \cdot \mathbf{K} \cdot \left(\vec{\nabla} (\mu^f - \mu_0^f) + \frac{n^i}{n^f} \vec{\nabla} (\mu^i - \mu_0^i) \right) = 0, \quad (19)$$

$$\frac{\partial n^i}{\partial t} + \vec{\nabla} \cdot (n^i \vec{v}^f) = \vec{\nabla} \cdot \mathbf{D}^i \cdot \vec{\nabla} (\mu^i - \mu_0^i). \quad (20)$$

2.2. DIFFUSIONAL MODEL FOR WATER VAPOR TRANSPORT IN AIR

Vapor diffusion through stagnant air can be derived from the mass balance of water vapor in air and the constitutive equation for water vapor transport

$$\frac{1}{R^v T} \frac{\partial p^v}{\partial t} - \frac{1}{R^v T} \vec{\nabla} \cdot (\mathbf{D}^v \cdot \vec{\nabla} p^v) = 0, \quad (21)$$

where R^v is the specific gas constant of water vapor, T the absolute temperature, $\mathbf{D}^v$ the diffusion tensor of water vapor in air, and p^v the water vapor pressure. The introduction of vapor convection by air motion is beyond the scope of this work.

2.3. BOUNDARY CONDITIONS

To arrive at a unique solution, specification of boundary conditions at the material surface is required. For the solid component it is possible to prescribe either the displacement, or the effective stress at the boundary.

For the fluid component either the molar chemical potential of the fluid, or the fluid flux can be prescribed.

When the triphasic mixture is in contact with air, continuity of the molar chemical potential of fluid across the skin–air boundary requires that the molar chemical potential of the water vapor at the skin surface equals the chemical potential of the fluid just beneath the skin surface

$$\bar{\mu}^v = \bar{\mu}^f = \mu^f \bar{V}^f, \quad (22)$$

where $\bar{\mu}^{v,f}$ is the molar chemical potential (vapor, water), μ^f the chemical potential of water, and $\bar{V}^f$ the molar volume of water. The molar chemical potential of water vapor is

$$\bar{\mu}^v = \bar{\mu}_0^v + RT \ln \left(\frac{p^v}{p_0^v} \right), \quad (23)$$

where p^v is the water vapor pressure, and p_0^v the water vapor pressure in a reference state (saturated state). Because the medium in contact with the triphasic mixture is not necessarily incompressible, the molar chemical potential is used instead of the volumetric chemical potential.

Combination of these two equations yields

$$\mu^f = p - \pi + \psi = \frac{RT}{\bar{V}^f} \ln \left(\frac{p^v}{p_0^v} \right). \quad (24)$$

If no condensation occurs, Equation (24) states that the chemical potential of the fluid at the skin surface (μ^f) lowers the actual vapor pressure at the skin surface (p^v) as compared to the saturated vapor pressure (p_0^v). In this study we do not consider the case of condensation, as perspiration usually does not result in condensation phenomena.

For the ionic component either the chemical potential of the ions or the ion flux can be prescribed.

2.4. NUMERICAL IMPLEMENTATION

To solve the set of equations for an arbitrary geometry and arbitrary boundary conditions, the finite element method is used. The present triphasic implementation is an adapted version of the implementation of Snijders *et al.* (1995) in the finite element code DIANA (Borst *et al.*, 1985). Time discretization is achieved by an implicit Euler integration scheme. In simulations physically nonlinear behavior as well as geometrical nonlinearity is included. The resulting nonlinear system of equations is solved by a regular Newton–Raphson technique and Gauss decomposition (DIANA, 1996).

3. Triphasic Material Parameters of Skin

Before simulation results can be obtained a constitutive equation for the effective Cauchy stress and the material parameters have to be specified.

For simplicity we focus on purely elastic models, but time dependent behavior of the skin is still possible due to the presence of the fluid in the triphasic model. An isotropic linear relation between the second Piola–Kirchhoff stress tensor ($\mathbf{S}$) and the Green–Lagrange strain tensor ($\mathbf{E}$) is used. Although more advanced models of stress–strain relationship are available in the literature, they all focus on in-plane properties and/or deformation under incompressible conditions, while we are in the need of properties perpendicular to the skin surface and under changing volume conditions. Considering that, in the context of this paper, we verify the model for one-dimensional conditions only, the assumption of isotropy is trivial. For small deformations a linear $\mathbf{S} - \mathbf{E}$ relation is identical to Hooke’s law.

3.1. POROSITY

The water volume fraction (or porosity (n^f)) varies over the depth of the skin and depends on the environmental conditions. Warner *et al.* (1988) measured the water concentration profile across rapidly frozen epidermis. The porosity in the triphasic theory is not the same as the total water content measured by Warner *et al.* Based on the experiments of Anderson *et al.* (1973) it is assumed that the intracellular water surrounded by the cell envelope (a water tight cell membrane through which water cannot pass easily) is part of the solid. Combination of measurements of Hansen and Yellin (1972), Anderson *et al.* (1973), and Warner *et al.* (1988) results in a porosity profile as a function of depth from the skin surface. This profile ranges from $n^f = 0.13$ at the skin surface to $n^f = 0.35$ in the deeper layers of the skin

$$n^f = 0.35 - \frac{1}{(8.04 \times 10^3 X + 1.05)^{28.27}}, \quad (25)$$

where X is the distance from the surface in the reference configuration.

The porosity is a function of the deformation. Assuming intrinsic incompressibility of the skin’s solid and fluid components, the porosity in the deformed state can be written in terms of the porosity in a reference state (n_0^f) and the determinant of the deformation gradient tensor J

$$n^f = 1 - \frac{1 - n_0^f}{J}. \quad (26)$$

3.2. MATRIC POTENTIAL

The matric potential arises from the interaction (e.g. capillary forces and surface adsorption) of water with the solid matrix. These forces vary with water content. Due to the interaction between solid, water, and ions, the water vapor pressure

Figure 1. Relation between chemical potential and porosity from three different references: (Anderson *et al.*, 1973; Spencer *et al.*, 1975; El-Shimi and Princen, 1978).

above the skin (p^v) is lower than the water vapor pressure above pure water (p_0^v) (Marshall, 1988). The chemical potential of the water at the surface of the skin is given by Equation (24). Measuring the vapor pressure above skin provides a means for examining the interaction between ions, water, and solid. Several authors determined the water content (porosity) of human stratum corneum as a function of ambient relative humidity (Anderson *et al.*, 1973; Spencer *et al.*, 1975; El-Shimi and Princen, 1978). Their data are used to obtain a relationship between fluid chemical potential and porosity. The following relation between matric potential and porosity is derived from their data

$$\mu^f = 9.84 \times 10^{10} (e^{9.7 \times 10^{-5}} - e^{9.7 \times 10^{-5}/n^f}). \quad (27)$$

Figure 1 shows the data measured by the cited authors and the fit according to Equation (27).

The individual contributions of the fluid pressure, the osmotic pressure and the matric potential to the chemical potential are difficult to obtain. The left ventricle pressure is in the order of 0.013 MPa (Guyton and Hall, 1996), the pressure in the intervertebral disco measured by Nachemson (1981) is in the order of 0.3 MPa. These values illustrate the usual order of magnitude of physiological pressures. Assuming a physiological salt concentration of 0.15 M NaCl yields an osmotic pressure of 0.7 MPa. Especially at low porosity these values are negligible as compared to the value of the matric potential. Therefore, as a first approximation, we accept that the matric potential is the only significant contributor to the chemical potential.

3.3. YOUNG'S MODULUS

Literature values for the Young's modulus of the dermis range in orders of magnitude (10^3 – 10^6 N/m²). The great variety in data is probably due to differences in

test region, test method etc. Several authors found that changes in relative ambient humidity affect mechanical properties of the skin. Park and Baddiel (1972) measured *in vitro* a Young's modulus profile of stratum corneum as a function of relative humidity. This profile can be fitted with an exponential function

$$E = 0.84e^{(-(RH/100+0.5)^{4.8}+22.0)}. \quad (28)$$

The relationship between water content and relative humidity (Anderson *et al.*, 1973; Spencer *et al.*, 1975; El-Shimi and Princen, 1978) is used to obtain Young's modulus as a function of the water content (porosity). Combination of this fit with the porosity profile in the skin reveals the Young's modulus of the epidermis as a function of skin depth.

3.4. HYDRAULIC PERMEABILITY

Swabb *et al.* (1974) reported a value of $6.4 \times 10^{-15} \text{ m}^4/(\text{Ns})$ for the hydraulic permeability of rat stratum corneum with a confidence limit of 25%. The hydraulic permeability of a soft hydrated tissue like cartilage lies between 10^{-14} and $10^{-16} \text{ m}^4/(\text{Ns})$ (Maroudas, 1975). Oomens (1985) used a biphasic model to fit the hydraulic permeability of pig skin. It equals $3.5 \times 10^{-14} \text{ m}^4/(\text{Ns})$ in the unstrained state.

We performed hydraulic permeability measurements on rat skin. A detailed description of these measurements is given in Kemenade (1998). The hydraulic permeability of four different samples ranged about a factor five from 2.3×10^{-15} to $1.2 \times 10^{-14} \text{ m}^4/(\text{Ns})$. It is difficult to check reproducibility because of deterioration of the skin samples. Measurements on synthetic porous gels (Oomens *et al.*, 1995) show a good reproducibility of the hydraulic permeability measurement with the capillary method used.

The skin consists of several layers with different properties. The upper layer of the skin, especially the stratum corneum is thought to be very impermeable for water flow. In our measurements intact skin samples are used which means that an overall permeability coefficient is obtained. In the measurement set-up the skin sample is surrounded by a physiological salt solution which results in hydration of the skin, especially of the stratum corneum. Because hydration dramatically increases the hydraulic permeability, the effect of the impermeable dry outer stratum corneum is not visible in the experiment. It is, therefore, assumed that the mean of the measured permeability coefficients ($6 \times 10^{-15} \text{ m}^4/(\text{Ns})$) corresponds with the hydraulic permeability coefficient of the dermis.

Maroudas (1975) measured the hydraulic permeability of cartilage as a function of hydration. She found a 15 times higher permeability at 100% hydration as compared to 50% hydration. Fatt (1968) measured the hydraulic permeability of corneal stroma as a function of hydration. He finds a 40 times higher permeability for 5 as compared to 2 g water per gram dry weight. It is, therefore, believed that the permeability of the stratum corneum with its very low water content is only a small fraction of the permeability measured in the experiment.

In steady state the fluid flow through the skin equals the transepidermal water loss. Fluid flow is described by Darcy's law

$$n^f(\vec{v}^f - \vec{v}^s) = -\mathbf{K} \cdot \vec{\nabla} \mu^f. \quad (29)$$

Multiplying this equation by the water density yields the mass flux of water through the skin. This flux is normally 5–10 g/(m²h). A fit of the chemical potential as a function of the porosity is given by Equation (27). Combining this equation with the porosity profile across the skin yields the gradient in the chemical potential across the skin. This gradient combined with a water mass flow of 10 g/(m²h) enables us to estimate the hydraulic permeability profile across the skin.

3.5. DIFFUSION TENSOR

Mackie and Meares (1955) derived a formula which relates the diffusion coefficient in a porous medium to that in a free solution

$$D = D_{\text{free}} \frac{(n^f)^2}{(2 - n^f)^2}, \quad (30)$$

where D_{free} is the diffusion coefficient in a free solution. In the triphasic model diffusion of positive and negative ions is coupled into diffusion of a neutral salt. For the diffusion coefficient of the salt the harmonic average of the diffusion coefficients of the separate ions is used. This results in a $D_{\text{free}}^{\text{NaCl}}$ of 1.61×10^{-9} m²/s.

As the porosity in the skin varies from about 0.13 to 0.35, the diffusion coefficient in skin is 0.005–0.06 times 1.61×10^{-9} m²/s. For simplicity one diffusion coefficient (based on $n^f = 0.25$) of 3.3×10^{-11} m²/s is used throughout the whole skin.

4. Methods

4.1. WATER LOSS MEASUREMENTS

We investigated transepidermal water loss and skin hydration on the volar forearm under varying ambient conditions. In experiment 1, humidity changed once about halfway the experiment (three subjects). In experiment 2, several cycles of relative humidity variation were applied by manually changing the room's settings (six subjects).

Nine healthy volunteers without skin diseases (aged 22–28 years) took part. Informed consent was obtained from all participants. Subjects were requested not to use any moisturizers or body lotions on the day of the measurement. Subjects were preconditioned by 30 min of rest.

Transepidermal water loss was measured using the TEWAmeter[®]. This instrument is based on the open chamber system with two humidity and temperature

sensors which measure the water evaporation gradient at the surface of the skin. The Corneometer CM820® (Courage and Khazaka) was used to measure skin hydration. This instrument measures the electrical capacitance of the top 30 μm of the skin in arbitrary units. Relative humidity and ambient temperature were measured with a thermo-hygrometer (RH82, Omega Technologies Company, Stamford, U.S.A.).

4.2. NUMERICAL SIMULATION

Because the experiment is performed on the volar forearm, the dimensions of this anatomical site are used in the finite element analysis. The mean thickness of the stratum corneum of the forearm is 13 μm (Holbrook and Odland, 1974). The thickness of the epidermis (including the stratum corneum) is 60 μm (Whitton and Everall, 1973). Considering the experimental evidence that the skin barrier is localized in the outermost layers of the skin, sink conditions are assumed at the level of the subpapillary plexus of blood vessels (located at 10% of the thickness of the dermis). This means that 0.2 mm of the dermis is modeled. Eight-node quadrilateral plain strain elements (element type CQ16E, (DIANA, 1996)) with eight displacements and four chemical potentials degree of freedom are used. Due to the strongly non-homogeneous character of skin, especially in the outer layer, the finite element mesh is very fine at the skin surface. The skin mesh consists of 98 skin elements with 591 nodes, ranging from a thickness of 0.1–7 μm . For simplicity and because of the configuration of the experimental set-up, strains parallel to the skin's surface are set to zero. Hence, the mesh consists of one element in transversal direction (width 0.2 mm).

The thickness of the air layer is chosen such that the same vapor pressure gradient as used in the transepidermal water loss measurements is simulated. Therefore, the air mesh is 13 mm high and consists of 13 user defined 8-node quadrilateral elements of equal size.

The skin surface is in contact with air. Thirteen air elements are added at the skin surface. At the transition between skin and air, the water vapor chemical potential equals the chemical potential of the water at the skin surface. This means that the mass flux of water leaving the skin equals the mass flux of vapor entering the air.

At the top of the air mesh (13 mm above the skin surface), a vapor chemical potential depending on the experimental ambient conditions is prescribed. The smooth cycles of relative humidity changes in experiment 2 are approached with piecewise linear relative humidity changes. At the lower end of the mesh the boundary conditions are: vanishing displacements and chemical potentials of water and ions consistent with contact with the physiological salt concentration and pressure of capillary blood.

Table I. Transepidermal water loss $\text{g}/(\text{m}^2 \text{ h})$ from intact skin for different material parameter profiles

	$0.5 * K$	$1.0 * K$	$2.0 * K$
$1.0 * E$	13.0	25.6	49.4
$2.0 * E$	8.8	17.3	33.7

5. Results

5.1. STEADY STATE WATER LOSS: PARAMETER VARIATION

In literature a great number of basal transepidermal water loss measurements on the volar forearm can be found ranging from about 3 to $12 \text{ g}/\text{m}^2 \text{ h}$.

Because of the uncertainty in some material parameters, a parametric study in which the hydraulic permeability and the Young's modulus vary, is performed. The initial hydraulic permeability profile across the skin is multiplied by a factor 0.5 and 2.0. The Young's modulus profile is multiplied by a factor 2.0. Table I summarizes the results of these simulations.

The results of the parameter variation clearly show the influence of the hydraulic permeability and the Young's modulus. An increase in permeability (decrease of resistance against fluid flow) results in an increase in transepidermal water loss. An increase in Young's modulus results in a decrease in transepidermal water loss. If the Young's modulus is multiplied by a factor 0.5, the simulation does not converge because the deformations become too large.

The hydraulic permeability at the surface of the skin is very small, resulting in a large negative value of the chemical potential at the skin surface. This large negative value reduces the vapor pressure at the skin surface in such away that realistic transepidermal water loss values are simulated. A steep gradient in the chemical potential of the fluid exists across the outer layers of the skin. Figure 2 shows

Figure 2. Left: Chemical potential of the fluid. Right: Different contributions of pressure potential (p), osmotic potential (π), and matric potential (ψ) to the chemical potential of the fluid.

the different contributions of the pressure, osmotic, and matric potential to the chemical potential of the fluid.

5.2. WATER LOSS UNDER VARYING AMBIENT RELATIVE HUMIDITY

In experiment 1, at $t = 60$ min, the relative humidity changes from 53 to 89%. The simulated and measured transepidermal water loss for 1 subject are presented in Figure 3.

The change of relative humidity from 53 to 89% results for both measurement and simulation in an immediate decrease in transepidermal water loss followed by a slow recovery of water loss to values lower than the original value. The other measured subjects showed the same pattern.

In the simulation the skin surface porosity change almost immediately after a change in relative humidity. The porosity of deeper layers of the stratum corneum reacts more slowly. The time constant associated with the change in porosity of the deeper layers is comparable to the time constant of the recovery of transepidermal water loss. The measured skin hydration gradually increased after increasing the humidity to 89% (data not shown).

In experiment 2, the relative humidity changes in time were achieved by manually changing the room settings. Therefore, all six subjects underwent different relative humidity variation and it is not possible to group the results. However, the reactions of transepidermal water loss on relative humidity changes followed the same pattern. In the upper part of Figure 4 typical patterns of relative humidity changes in time and corresponding transepidermal water loss are shown.

After the first change in relative humidity from 67% to about 100% transepidermal water loss immediately decreases followed by a recovery after humidity is decreased again. After 70 min the transepidermal water loss decreases while relative humidity also decreases. After long measurement times subjects tend to get cold hands and arms, which probably explains the lower transepidermal water loss, under decreasing relative humidity.

Figure 3. Simulated transepidermal water loss (—) and measured transepidermal water loss ($\square$) (experiment 1: $t = 0$ –59 min: RH = 53%; $t = 60$ –90 min: RH = 89%).

Figure 4. (a) Transepidermal water loss (□) and relative humidity (+) as a function of time. (b) Simulated transepidermal water loss (—) and measured transepidermal water loss (□) (experiment 2, subject 8).

Figure 4(b) shows the simulated and measured transepidermal water loss.

Comparison of measured and simulated water loss shows that the trends and amplitude in water loss are consistent. High relative humidity corresponds with low transepidermal water loss and vice versa, both in experiment and simulation.

6. Discussion

6.1. STEADY STATE WATER LOSS

The influence of Young's modulus on simulated transepidermal water loss (Table I) emphasizes the importance of the presence of a deformable solid in the model. The hydraulic permeability at the surface of the skin is very small, resulting in a large negative value of the chemical potential at the skin surface. Figure 2 shows that the contribution of the osmotic potential is negligible, and that the contribution of the matric potential especially in the outermost layers of the skin is very important. If the chemical potential gradient is due to a gradient in pressure potential alone, then very high effective stresses and unrealistically high deformations would be produced. Therefore, the matric potential is essential in the modeling of fluid flow

through the skin. Except in the outermost layers of the skin, a positive pressure potential is simulated. This results in positive effective stresses. Thus a tensile prestress is predicted which is consistent with the findings of Langer (1861). This positive pressure potential is probably overestimated, because the matric potential used in the deeper layers of the skin is based on measurements on stratum corneum. Moreover, we assumed a porosity of 0.35 in the deeper layers, resulting in a solid fraction of 0.65. In this large solid fraction with adsorptive properties the intracellular water is incorporated. However, this intracellular water fraction does not have adsorptive properties. The negative pressures developed in the outer layers (X_i 0.005 mm) of the stratum corneum are associated with the traction exerted on the water by the relatively dry environment of the air.

A more accurate fitting of the model to the experimental data shown in Figure 4, is probably not useful, as the present experimental techniques for measuring transepidermal water loss are reliable only in a relative sense, while absolute values are subject to double digit percentage uncertainties.

6.2. WATER LOSS UNDER VARYING AMBIENT RELATIVE HUMIDITY

In experiment 1, the initial decrease in transepidermal water loss is somewhat higher in the simulation as compared to the measurement, which might be explained by the time necessary to obtain a transepidermal water loss measurement (60 s) as compared to the time step in the simulation which is 20 s. The simulated decrease in transepidermal water loss at 61 min equals the measured decrease at 61 min. The resulting difference in transepidermal water loss between an ambient relative humidity of 53 and 89% at large values of t is both in experiment and simulation about 1 g/m² h.

The recovery can probably be explained by a slow increase in skin hydration as a result of the increasing relative humidity. Because this results in an increase in skin vapor pressure, transepidermal water loss increases. The fact that the time constant of the simulated change in porosity of the deeper layers of the stratum corneum is comparable to the time constant of the recovery of transepidermal water loss confirms this explanation. This recovery effect is also seen in (*in vitro*) measurements of Hilliard Jr. and Dorogi (1989), where after lowering the relative humidity, transepidermal water loss immediately increases and afterwards decreases to a transepidermal water loss higher than the original value.

In Figure 4 it can be seen that the simulated water loss especially in the second and third cycle of relative humidity variation varies somewhat less than the measured water loss. This could be due to the fact that the skin hydration (indicated by the porosity) in the model changes immediately after relative humidity variation while in experiments the change in skin hydration is much slower than the variation in relative humidity. The opposite trends of transepidermal water loss and relative humidity at t larger than 70 min may be explained by cooling down of the subjects arm, resulting in a lower transepidermal water loss.

7. Conclusions

We proposed a triphasic skin model coupled with air to describe water transport through the skin. As is usual in porous media theories we included hydration forces in an averaged sense as gradients of a matric potential (Nitao and Bear, 1996).

The simulations were compared with *in vivo* measurements, and it was shown that the model is able to simulate the measured decrease and the subsequent recovery of the transepidermal water loss resulting from an increase in relative humidity. The simulations show that the time constant of the recovery effect is comparable to the time constant of the change in the porosity of the surface layers of the skin, confirming the association between the recovery and a change in skin hydration. To the best of our knowledge this study presents the first *in vivo* verification of a triphasic model of a biological tissue.

In conclusion, the theory is capable of simulating initial decrease and subsequent increase of transepidermal water loss following an increase of ambient relative humidity. The analysis shows that the matric potential accounting for the fluid–solid interaction is essential in skin barrier function. We also think that the mechanics of the solid component is essential for the description of the recovery effect.

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