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Computational modeling of skin: Using stress profiles as predictor for tissue necrosis in reconstructive surgery

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ABSTRACT

Local skin flaps have revolutionized reconstructive surgery. Mechanical loading is critical for flap survival: excessive tissue tension reduces blood supply and induces tissue necrosis. However, skin flaps have never been analyzed mechanically. Here we explore the stress profiles of two common flap designs, direct advancement flaps and double back-cut flaps. Our simulations predict a direct correlation between regions of maximum stress and tissue necrosis. This suggests that elevated stress could serve as predictor for flap failure. Our model is a promising step towards computer-guided reconstructive surgery with the goal to minimize stress, accelerate healing, minimize scarring, and optimize tissue use.

1. Motivation

Tissue expansion is a reconstructive surgical technique that has established itself as a reliable method to correct birth defects, burns injuries, and regions of tumor removal [20]. The procedure was introduced fifty years ago, when the first implanted balloon was successively inflated to grow skin in situ to resurface a damaged ear [43]. Since then, skin expansion has been eagerly adopted by clinicians and has revolutionized the field of plastic and reconstructive surgery. Skin expansion creates new skin with the same mechanical, hair bearing, color, and texture characteristics as the surrounding tissue, which is ideal from an aesthetic point of view [52].

In current clinical practice, a medical device that resembles a fluid-filled balloon is implanted subcutaneously and inflated gradually over a period of weeks. Skin responds to controlled overstretch by growing in area to recover its homeostatic state [13,53]. When the device is removed, the newly grown skin is made available for reconstructive purposes through a process called flap design [27]. Careful planning of flap design is key to successful defect repair [11]. Yet, since most efforts of pre-operative planning focus exclusively on kinematic issues [25], flap failure remains a major complication in plastic and reconstructive surgery [38]. Obviously, the plain survival of the flap has the top-most priority. However, sub-optimal flap design can also compromise healing and trigger keloid formation and hypertrophic scarring [60].

The causes of flap complications are numerous, but mechanical factors are hypothesized to play a central role [45]: excessive tissue tension can create compression of the pedicle, compromise blood supply, and trigger tissue necrosis [58]. Fig. 1 illustrates a clinical example of flap necrosis as a result of excessive tissue tension. To resurface a giant birth defect, two tissue expanders were inserted into the right leg. After the expanders were fully inflated, two flaps were designed from the newly grown tissue and advanced to cover the excised defect. Two days after flap advancement, the distal leg displayed severe flap necrosis. The necrotic tissue had to be removed in a secondary procedure to promote healing. There is a general agreement that mechanical loading greatly influences the success of the reconstructive procedure. Surprisingly, despite all these evidences, the design of tissue flaps has never been analyzed from a mechanistic point of view.

A major cause for recurring flap failure arises from the difficulty to determine tissue stress in vivo, a problem that spans all surgical disciplines [62]. Even if mechanical cues are recognized as critical factors, measuring the stress profile in the clinical setting is virtually impossible. Computational models offer an excellent alternative to predict tissue stresses, provided the underlying constitutive models are appropriately calibrated [10,39]. The past two decades have

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Fig. 1. Distal flap necrosis induced by excessive tissue tension. (a) Giant congenital pigmented nevus on the right lower leg of a two-year old boy. (b) Tissue expanders inserted on the lateral and medial sides of the upper thigh, filled to capacity. (c) Lateral leg two days after resurfacing with rotation-advancement flap from expanded lateral thigh tissue. (d) Medial leg two days after resurfacing with direct-advancement flap from expanded medial thigh tissue demonstrating distal flap necrosis. (e) Necrotic tissue of distal flap debrided to promote healing by secondary intention. (f) Medial leg two months after flap advancement, with healing of distal aspect of flap by secondary intention.

seen remarkable advances in computational modeling as a useful tool in surgery planning and treatment optimization [61,64]. For skin, computational modeling has only recently received increased attention [31,47].

Skin is a remarkable tissue and the largest organ of our body. In the adult, it covers an area of approximately 2 m^2 and weights about 3.5 kg. The main functions of skin are to protect our interior body from the outside world, to regulate temperature and water exchange, and insulate our internal machinery from harmful substances and solar radiation [49]. Additionally, skin is densely packed with neural receptors that make it our largest sensor, the one responsible for our tactile sense [12]. Although it is a very thin membrane, only 8–14 mm thick, skin consists of three distinct layers [66]. The outermost layer, the epidermis, consists of several sublayers of cells stacked on top of one another. Cells in the inner epidermis undergo constant mitosis to continuously regenerate the epidermis; cells in the outer epidermis are dead [54]. The inner epidermis is connected to the dermis, the main load bearing layer of skin [55]. It consists of a water-based matrix that serves as a scaffold for its collagen and elastin fiber network. The hypodermis, which lies underneath the dermis, consists primarily of fat. Its function is to anchor the skin to the underlying bone and muscle tissue.

From a mechanical point of view, skin is a unique material [37]. Like in many other collagenous tissues, the collagenous network of the dermis is the key contributor to its mechanical function [33]. Skin is highly anisotropic. Its collagen bundles form an interwoven network aligned with a preferred microstructural direction [34]. This was first summarized more than a century ago in a comprehensive map of collagen orientations across the human body [32]. Collagen fibers undulation creates a characteristic non-linear response with locking stretches above which skin stiffens significantly [4,30]. It is not surprising that determining a suitable material model for skin and accurately identifying its parameters is not a trivial task [18].

One of the pioneering constitutive models for skin used biaxial tests of rabbit skin and fitted the data to an exponential strain energy function [57]. A following, more advanced model introduced a microstructurally motivated strain energy function with separate contributions for matrix and fibers [35]. This model was recently calibrated using pig skin experiments [26]. Latterly, an invariant formulation, initially designed to model the arterial wall, was adopted to model skin [23,44]. Several groups have tested skin in uniaxial tension [57], biaxial stretch [33], oscillatory shear [31], indentation [46], or general three-dimensional loading scenarios [16] to determine its material parameters. However, the underlying constitutive models vary from one group to another: Some assume isotropy as a pragmatic simplification [2,14,22], others claim that an accurate model should incorporate the anisotropic response [15,56].

In flap design, when stress profiles are critical for pre-operative planning, we need to carefully select the constitutive model and its material parameters. An accurate model of the skin's acute behavior will improve our understanding of the chronic adaptation in response to a surgical procedure. However, determining the stress distribution during flap design poses several challenges: following tissue expansion, the newly grown skin takes a complex three-dimensional dome-like shape [25], which needs to be stretched to resurface planar two regions [19]. During this process, skin undergoes extreme deformations including kinematic and constitutive nonlinearities, anisotropy, and stretch locking. Accurately simulating these characteristics presents tremendous opportunities for computational structural mechanics. The objective of this manuscript is to illustrate the use of computational structural mechanics to identify regions of maximum stress as potential indicators for tissue necrosis and clinical complication.

The manuscript is organized as follows. In Section 2, we summarize the constitutive equations used for the continuum mechanics modeling of skin. To model skin we use a microstructurally motivated, invariant-based, hyperelastic constitutive model. In Section 3, we present the finite element formulation of the problem. In Section 4, we investigate the two most common flap designs used following skin expansion, the direct advancement flap and the double back-cut flap. In Section 5, we draw important conclusions and discuss the potential of the proposed method to improve surgical outcomes in the plastic and reconstructive surgery field.

2. Governing equations

We begin by briefly summarizing the governing equations to model skin. Let $\mathcal{B} \subset \mathbb{R}^3$ be the reference configuration of a body that occupies $\mathcal{S} \subset \mathbb{R}^3$ in the deformed state. Material points $\mathbf{X} \in \mathcal{B}$ are mapped to points $\mathbf{x} \in \mathcal{S}$ trough the deformation map φ. We introduce the deformation gradient $\boldsymbol{F} = \nabla\varphi$ as the spatial gradient of the deformation map, which relates the tangent space of the reference configuration to the tangent space of the current configuration. Due to the incompressible nature of skin, we multiplicatively decompose $\boldsymbol{F}$ into a volumetric and an isochoric part,

$$\boldsymbol{F} = \boldsymbol{F}^{\mathrm{vol}} \cdot \bar{\boldsymbol{F}} \text{ with } \boldsymbol{F}^{\mathrm{vol}} = J^{1/3}\boldsymbol{I} \text{ and } \bar{\boldsymbol{F}} = J^{-1/3}\boldsymbol{F}. \tag{1}$$

Here, $J = \det(\boldsymbol{F})$ denotes the Jacobian, which characterizes the mapping of volume elements. Incompressibility implies that $\det(\bar{\boldsymbol{F}}) = 1$ and, consequently, $J = \det(\boldsymbol{F}) = \det(\boldsymbol{F}^{\mathrm{vol}}) \geq 0$. Local deformation is represented through the right Cauchy–Green deformation tensor $\boldsymbol{C}$ and its isochoric part $\bar{\boldsymbol{C}}$,

$$\boldsymbol{C} = \boldsymbol{F}^{\mathrm{t}} \cdot \boldsymbol{F} = J^{2/3}\bar{\boldsymbol{F}} \quad \text{with} \quad \bar{\boldsymbol{C}} = \bar{\boldsymbol{F}}^{\mathrm{t}} \cdot \bar{\boldsymbol{F}}. \tag{2}$$