Progress in Biophysics and Molecular Biology 103 (2010) 208–216

Contents lists available at ScienceDirect

Progress in Biophysics and Molecular Biology

journal homepage: www.elsevier.com/locate/pbiomolbio

Original research

Simulating plastic surgery: From human skin tensile tests, through hyperelastic finite element models to real-time haptics

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ARTICLE INFO

Article history:
Available online 30 September 2010

Keywords:
Hyperelasticity
Explicit finite element analysis
Human skin
Open surgery simulation
Programmable GPU
Haptic and force feedback

ABSTRACT

In this paper, we provide a summary of a number of experiments we conducted to arrive at a prototype real-time simulator for plastic surgical interventions such as skin flap repair and inguinal herniotomy. We started our research with a series of in-vitro tensile stress tests on human skin, harvested from female patients undergoing plastic reconstructive surgery. We then used the acquired stress–strain data to fit hyperelastic models. Three models were considered: General Polynomial, Reduced Polynomial and Ogden. Only Reduced Polynomial models were found to be stable, hence they progressed to the next stage to be used in an explicit finite element model aimed at real-time performance in conjunction with a haptic feedback device. A total Lagrangian formulation with the half-step central difference method was employed to integrate the dynamic equation of motion of the mesh. The mesh was integrated into two versions of a real-time skin simulator: a single-threaded version running on a computer's main central processing unit and a multi-threaded version running on the computer's graphics card. The latter was achieved by exploiting recent advances in programmable graphics technology.

1. Introduction

Our research on plastic surgery simulation was initiated by performing in-vitro experiments on human skin, followed by fitting hyperelastic models to the data from the earlier experiment. These models were then subsequently used in an explicit finite element (FE) model which was accelerated on the graphics processing unit (GPU) of the video-adapter card for real-time interaction. Finally, the accuracy of the simulator was tested by comparing it to the original models and the established finite element software ABAQUS, whilst real-time performance was assessed by monitoring the update rate during interactions. The remainder of this section will give a brief overview on the literature of relevant experiments on the mechanical behaviour of human skin and soft tissue simulation for surgical interventions.

1.1. Human skin testing

Human skin, when deformed, displays both time-dependent or "viscoelastic" deformation and time-independent or "hyperelastic" deformation. The work presented in this paper originates from the application of computerised simulation for plastic surgical interventions such as skin-flap repair for facial reconstruction, where skin is suddenly stretched to a new location in a short period of time. It is for this reason that we only investigate the hyperelastic component of human skin deformation[1]. The mechanical behaviour of human skin varies across the body and between individuals. Skin is anisotropic and locally orthotropic. This property was demonstrated by Karl Langer in 1861 (Langer, 1861), who noticed that circular holes punched in the skin of cadavers relaxed into elliptical shapes. He drew lines (subsequently called Langer lines) through the principal axes of the elliptic cut-outs which approximate the direction of collagen bundles within the skin. Later, Borges (Borges, 1984) described Relaxed Skin Tension Lines (RSTLs), which are more precise than Langer lines as they are determined from live humans. RSTLs follow furrows formed when the skin is relaxed. Because skin's mechanical behaviour is affected by many factors, including the age, weight and lifestyle of individuals, only generalised models can be created. The situation is also compounded by the difficulty of collecting data for these skin models. Both in vivo and in vitro

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[1] The viscoelastic component is important when modelling the outcome of a plastic surgery intervention, e.g. how will the patient look after surgery, when replaced skin has settled to an equilibrium. This is not part of our current study.

0079-6107/$ – see front matter © 2010 Published by Elsevier Ltd.
doi:10.1016/j.pbiomolbio.2010.09.013

experiments have been conducted to determine skin's properties, and both have their advantages and disadvantages. In vivo testing allows tissues to be tested in their natural state. Proponents of this type of testing argue that skin's mechanical behaviour alters immediately when removed from the body and thus, in vitro testing does not reproduce the correct behaviour of skin. However, advocates of in vitro testing argue that it is extremely difficult to ascertain the boundary conditions of samples tested in vivo and to isolate their mechanical behaviour from the influence of surrounding tissues. Moreover, in vitro tests allow much larger strain evaluations as compared to in vivo tests on live humans. An in-depth literature review on both in vitro and in vivo experiments on human skin is beyond the scope of this paper. However, a good overview can be found in Gasson's thesis (Gasson, 2008).

Table 1 shows a summary of experimental values as reported by different authors on both in vivo and in vitro testing. It shows a wide range of values for Young's modulus, ranging from 4 kPa up to 18.8 MPa. This wide range of values is a result of skin's non-linear mechanical behaviour, which makes Young's modulus a somewhat questionable measurement of skin's elasticity, unless used to record the lowest and highest elastic moduli. The difference in values is also due to the wide range of elastic models used by the researchers which includes: linear elastic, viscoelastic (using spring and dashpot elements) and hyperelastic.

1.2. Surgical simulation

The end goal of our research is a real-time plastic surgery simulation in an interactive virtual environment. Most surgical simulations require accurate soft tissues models which operate in real-time. The main methods for modelling deformable objects are mass-spring-damper (MSD) and finite element (FE) models. MSD meshes are computationally efficient (compared to FE meshes) and are simple to implement. MSD meshes are popular amongst implementers of surgical simulators and have been used extensively. Because they are based on a discrete model, they are not physically as accurate as the FE models. Although work has been done on improving their accuracy by researchers such as Bianchi et al. (2003, 2004), Gelder (1998) and many others, the more computationally expensive and accurate FEM (Finite Element Method) is preferred. Research, such as that by Cotin et al. (1999, 2000) has allowed linear elastic FE meshes to be used in real-time simulations. More recent work by Taylor et al. (2007, 2009) allows large non-linear dynamic explicit FE meshes to be used in real-time as required for modelling most biological soft tissues. Indeed, it is the massive advancement in graphics hardware technology that makes real-time surgical simulation of complex non-linear elastic deformable objects a reality. Whereas before, a vast number of expensive clusters of computers were required to perform non-linear dynamic explicit FE simulations in real-time, the programmability and processing power of modern GPUs (Graphics Processor Units) allow these simulations to be run on inexpensive consumer hardware found in high-end PCs. The non-linear dynamic explicit FEM is the preferred method for modelling the biomechanical behaviour of human skin in a real-time surgical simulator, where accurate stresses must be produced for realistic haptic/force feedback.

Table 1
Values for Young's modulus of human skin as recorded by various research teams. LEGEND: LE = Linear Elastic; HE = Hyperelastic; VE = Viscoelastic. RSTL = Relaxed Skin Tension Line.

Researchers	Young's mod.	Test	Model
Diridollou et al. (1998, 2000)	129 kPa ± 88 kPa	in vivo (aspiration)	Hookean LE
Hendriks et al. (2003)	56 kPa, 177.6 kPa	in vivo (aspiration)	Mooney–Rivlin HE
Khatyr et al. (2004)	657 kPa ‖to RSTL 130 kPa ⊥ to RSTL	in vivo (extension)	linear VE
Jachowicz et al. (2007)	7–33 kPa	in vivo (indentation)	linear VE
Leveque et al. (1980), Agache et al. (1980)	850 kPa (age > 30) 420 kPa (age > 30)	in vivo (torsion)	LE
Dunn et al. (Dunn and Silver, 1983)	1.586 MPa (47%) 3.164 MPa (62%)	in vitro	LE Linear VE
Silver et al. (Silver et al., 2001)	0.1 MPa 18.8 MPa	in vitro	VE

2. Methodology

2.1. In vitro experiments on human skin

The primary objective of the skin experiments is to produce stress–strain response data to find a suitable hyperelastic model of skin's mechanical behaviour during plastic surgery procedures such as skin-flap repair. Due to the relatively short time it takes to stretch the skin, the effect of time on the deformation is negligible, hence viscoelastic effects are not needed in the real-time finite element solution presented later. However, this does not imply that viscoelastic effects should be ignored as a whole. Indeed, the latter effects will be important to observe from the skin's stress–strain response during stretching, to evaluate the quality of experimental testing and in particular the clamping of the samples to avoid slippage.

From the previous section it appears that the majority of recently reported skin experiments have been conducted in vivo. Most of these experiments are performed non-invasively on the surface of the skin and test samples (volunteers) are readily available, which makes them particularly appealing. Researchers also comment that the material properties they find are more realistic than those found in vitro, as the skin is in its natural state. However, the boundary conditions during in vivo testing are very difficult to define, even when using some of the methods as suggested in Hendriks et al. (2003), Jacquet et al. (2008), Lim et al. (2008). In vitro testing has been chosen for the experiments described in this paper based on the assumption that experimental error due to deterioration of the skin following excision is bound to be less than that found during in vivo testing, where the boundary effects are complex and difficult to define. Also, skin flaps exhibit large strain displacements which better suits in vitro testing. We have reported an initial in vitro experiment on human skin in Gasson et al. (2008). There, we discussed problems encountered to clamp skin tissue samples securely. In this paper, we will not reiterate these issues but we will give a brief overview of the experimental setup and the improved clamping of skin tissue samples.

2.1.1. Skin Sample Preparation

An abdominal skin flap of a healthy middle-aged female following abdominoplasty was provided in a sealed container, suspended in a saline solution and kept in a cool environment at all times (temperature around 10 °C). The size of the skin flap was approximately 5″ by 3″ when unfolded. Tests were performed approximately 6 days after removal of the skin from the patient. Samples were cut into parallel strips using a razor blade and sharp knife by an experienced surgeon. Scissors were used to remove the subcutaneous fat from the back of both uni-axial and planar test samples, leaving the dermis and epidermis layers intact. As with