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Biomechanics for computer-assisted surgery

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1. Introduction

TIMC-IMAG Laboratory (http://www-timc.imag.fr) is a 250-people laboratory devoted to translational and fundamental research at the intersection between medicine, information science and technology (from applied mathematics to computer science and robotics). In that laboratory, since the 1980s, the computer-assisted medical intervention group has been developing devices to assist the physician or the surgeon in the successful execution of diagnostic or therapeutic gestures by minimising invasiveness whilst improving accuracy.

Computer-assisted surgery (CAS) is now a mature domain. Researchers, clinicians and industrial partners have developed CAS applications by building links with classical domains such as computer science, robotics, image processing and mathematics. Orthopaedics was the first clinical domain mainly addressed by the pioneer CAS applications (Taylor et al. 1996). The reason for this was probably that bones are the human body structures which were considered as the most easily includable into a CAS application: they were assumed to be rigid, i.e. with a fixed 3D geometry, they are strongly identifiable onto computed tomography (CT) exams, and their relative position during surgery is easily tractable by fixing rigid bodies onto their external surfaces (these rigid bodies being, for example, tracked with the use of an optical device, thus providing 'surgical navigation').

The connection to biomechanics (i.e. the mechanics of living tissues) is more recent. Biomechanicians were first asked to work onto CAS applications when orthopaedic surgeons were looking for tools able to predict risks of fractures in the case of prosthetic implants. In that case, bonny structures could no more be considered as rigid, but on the contrary had to be modelled as a deformable continuum with a non-homogeneous distribution of the internal stresses. For example, a patient-specific finite element (FE) model of the femur could be designed to estimate the internal stresses generated by a hip prosthesis and therefore to help limit fracture risks (Weinans et al. 1994). In these continuous biomechanical models, bones were usually considered as linear elastic material that underwent small deformations, which permitted easy calculation of numerical solutions.

More recently, CAS has addressed a larger spectrum of clinical domains such as cardiology, neurosurgery, urology or abdominal surgery. For these applications, biomechanics faces a new challenge since the involved tissues are required to move and be deformed by stress generated by clinical actions. Moreover, soft tissues are difficult to model accurately since they typically exhibit complex, time-dependent, nonlinear, inhomogeneous and anisotropic behaviours. Most of the corresponding biomechanical models need to include large deformation effects and visco-hyperelastic constitutive laws. Such models are very computationally demanding and are therefore limited to pre-operative use, since the simulations often require many minutes or hours to compute.

Our group did contribute to such pre-operative use of biomechanical models, for example, in the domain of orthognathic surgery (Chabanas et al. 2003), tongue cancer treatment (Buchaillard et al. 2007) or orbital surgery (Luboz et al. 2007).

More recently, we have addressed the new frontier that biomechanics is now facing with the development of CAS devices that can provide *intra-operative* assistance (Payan 2012). The underlying idea is to use patient-specific biomechanical models during surgery, i.e. in the operating theatre. In that case, three main challenges need to be solved to be compatible with the clinical constraints:

(1) Patient-specific models should be easily generated (not more than some minutes to elaborate such a model).
(2) Patient-specific constitutive equations of the soft tissues have to be estimated through *in vivo* experiments, some of them only being possible during surgery if the organs are not accessed pre-operatively (e.g. the brain tissues).

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(3) The implementation of the models should provide real-time (or at least interactive) numerical simulations.

2. Generation of patient-specific models

In order to face the time constraints for the design of patient-specific FE models, our group has proposed the *Mesh-Matching* algorithm (Couteau et al. 2000) followed by the *Mesh-Match-and-Repair* (MMRep) approach (Bucki et al. 2010). The idea consists in maintaining the advantages of a manual design of the FE mesh, while introducing an automatic mesh conformation process. The global algorithm is based on the following strategy:

(1) A FE 'generic' (or 'atlas') biomechanical model is manually built. This step is long, tedious, but is done once.
(2) Patient data are collected (ultrasound (US), CT or magnetic resonance imaging (MRI).
(3) The generic FE mesh is automatically conformed to patient morphology (extracted from the data), by means of a local elastic registration.
(4) The new patient mesh is regularised so that it can be used for FE analysis. This patient mesh has a topology similar to the generic mesh (same number of elements and same element types).

3. Estimation of the constitutive law

In order to estimate the *in vivo* constitutive behaviour of human soft tissues, our group has developed the light aspiration device for *in vivo* soft tissue characterisation (LASTIC) based on the pipette aspiration principle and consisting in measuring the tissue deformations induced by a negative pressure. It is built in a very compact metallic cylinder of 33 mm in height and 34 mm in diameter (Schiavone et al. 2010). The surgeon holds the instrument and establishes contact with the tissues surface while the device measures the negative pressures and displacement responses. The LASTIC device can undergo a full sterilisation, and the data processing is sufficiently fast to provide an interactive estimation of the soft tissues' constitutive equation. LASTIC has already been used to evaluate the constitutive behaviour of forearm skin, tongue (Schiavone et al. 2008) and brain tissues (Schiavone et al. 2009).

4. Interactive-time numerical simulations

Last but not the least is the constraint of a quasi-real-time computation of the simulations provided by the models, in order to be used during surgery. This can be ensured when tissue deformations are small (e.g. brain deformations during large skull openings; (Bucki et al. 2012)) but it is still challenging for models with large visco-hyperelastic frameworks (see, for example, Stavness et al. (2011)). This bottleneck will definitively need to be studied in the future.

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