

---

# Assessment of Biomechanical Behavior of Forearm Ligaments in Numerical Simulations

## Technical Report

---

**Authors :**

Noura Hamze

Fabio Carrillo

Philipp Färnstahl

Matthias Harders

This work is supported by the Swiss National Science Foundation, under grant number 325230L\_163308, and the Austrian Science Fund FWF, under grant number I 2545-N31.

# 1 Introduction

In this report, we present numerical simulations which aim at identifying the appropriate constitutive law for forearm ligaments. The present work constitutes subpart of a larger research project focusing on investigating new methods for computerized preoperative simulations of the upper extremities. It can be considered as a continuity of our two previous recent studies; the first one [4] proposes an automatic musculoskeletal modelling framework, and the second [1] investigates the properties of the forearm’s interosseous membrane (IOM) in cadaveric specimens.

We start with a brief description of our conducted reference studies. Then, we present our numerical simulation approach and detail the experimental settings. Further, we compare the results of the numerical simulations to ex-vivo experiments. Finally, we conclude with comments and recommendations.

## 2 Tensile reference studies

In [1], project partners have investigated the tensile properties of the IOM structures for five fresh frozen cadaver forearm specimens. Each specimen has been divided to three ligamentous sections, and the rest of the anatomical structures have been carefully resected. Firstly, all the morphological properties of the ligaments including their widths, thicknesses, and attachment sites have been measured and fed into our automatic modelling framework [4], and a set of 15 ligaments for the same specimens have been modelled with high accuracy ( $< 0.58$  mm average main square error and  $< 7$  mm Hausdorff distance error). Secondly, tensile ex-vivo experiments have been performed. For each ligament, the radius has been fixed, and a loading force has been applied on the corresponding segment of the ulna. The force direction was aligned to the main direction of the ligament’s fibers; a constant displacement rate of 5 mm/min has been applied, and the load was stopped once the ligament tissue started to yield. Applied force, displacements, yield force as well as the strains have been measured for every single ligament. Later, these measurements have served as a ground truth to validate our numerical simulations. In the following, we explain our biomechanical simulation approach.

## 3 Numerical ligament simulations

The ligamentous soft tissues are regarded as deformable objects which are subject to the laws of continuum mechanics. One numerical method to solve the governing equations of motion is the finite element method (FEM). In this theory, the continuum is partitioned into smaller disjoint cells called elements. In the present work, we use Lagrange tetrahedral elements, composed of four nodes, each with three degrees of freedom. The equation of motion is solved for the nodes of the element, and the values inside the element are obtained with an interpolation function defined for each node. Due to the rather slow motions, we consider the problem as quasi-static and only look for the configuration of the ligament at that equilibrium, disregarding the dynamic transient effects. Thus, the discrete equation to solve is  $\mathbf{f}(\mathbf{x}) = \mathbf{0}$  where  $\mathbf{x}$  and  $\mathbf{f}$  are respectively the position and the force vectors on the nodes of the tetrahedral elements. In the given configuration,  $\mathbf{f}$  is a non-linear function of the position of the nodes  $\mathbf{x}$ , and represents the sum of the internal and external forces. In order to solve the equation, we use the following first-order linearization at each time step :  $\mathbf{f}(\mathbf{x}+d\mathbf{x}) = \mathbf{f}(\mathbf{x}) + \mathbf{K}(\mathbf{x})d\mathbf{x}$  where the Jacobian matrix  $\mathbf{K}(\mathbf{x}) = \partial\mathbf{f}/\partial\mathbf{x}$  depends on the nodes’ position (also addressing preload within ligaments). Temporal integration is performed based on the implicit backward Euler scheme, iteratively until reaching equilibrium.

In order to define the tissue behavior, a constitutive law using the Neo-Hookean model is assigned to the ligaments. The boundary conditions are set using homogeneous Dirichlet conditions in the areas corresponding to the connection between a bone and a ligament. Those conditions are velocity projection constraints which prevents any movement of the ligament by adjusting the connection nodes to their initial position during every iteration in the solver.

## 4 Experimental setting

In order to mimic the ex-vivo tensile experimental setup, a group of 15 ligaments corresponding to the five specimens (3 per forearm) have been created by our framework [4]. The domain of each of them is modelled as a 3D tetrahedron mesh, and both the width and thickness values are set according to the measurements which are reported in [1]. Further, the Young’s modulus parameter which defines the stiffness of tissue has been

set individually for each ligament from the results obtained in the reference study. The Poisson's ratio which defines the compressibility is set 0.48 for all the ligaments as in the related literature. Afterwards, a dedicated simulation scene has been designed for each single ligament following the approach presented in Section 3.

An example of a simulation setup is illustrated in Figure 1. The ligament is rendered in red with the fixation constraints applied on the nodes which are located at the connection sites. Since the radius bone is kept intact and remains fixed during the simulation, it is sufficient to only keep its attachment constraints with the ligament at the nodal connection sites. As for the ulna, the bone has been cut to segments corresponding to the ligamentous component. This allows aligning the force with the main direction of the ligament which is defined by a vector pointing from the mid-point of the ligament's site on the radius to the opposing mid-point on the ulna. In the figure, an example of the ulna segment is rendered in yellow, with the applied force illustrated as a green arrow.

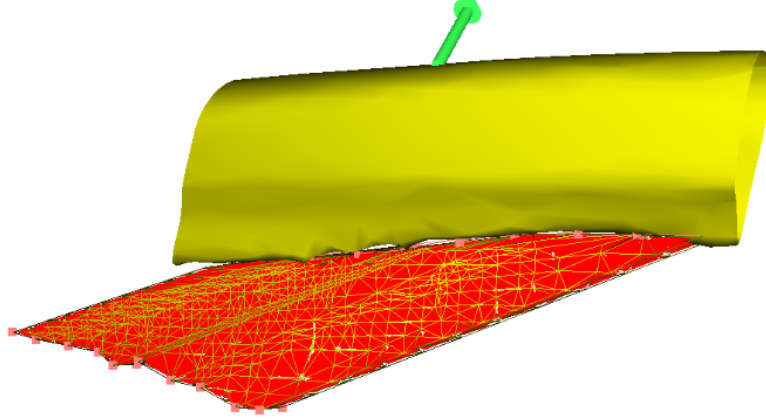


FIGURE 1 – An example of a simulation scene, rendered in SOFA [2].

Each simulation is associated with a force range starting from zero, incremented iteratively until reaching the yield force. The latter corresponds to the first peak force in the tensile experiments at which the ligament ramps to failure. The entire force is distributed on the ligament's nodes at the connection sites with the ulna's segment. Their directions are aligned with the main direction of the fibers. The loading forces will pull the ligament to a certain extent, and the simulation is considered to be completed (i.e. tissue failure) once the accumulated forces reach the provided yield force. Figure 2 depicts two states in one of the implemented simulations. Note that a preload of 0.5 N has been set inside the ligaments, corresponding to an internal tension force which exists in the IOM, and has been considered in the tensile experiments as well. The colors of volume elements correspond to the values of Von-Mises stresses computed during the simulation. The average time for a complete simulation varies between 20 and 40 sec on standard computing hardware, making them usable in a reasonable time.

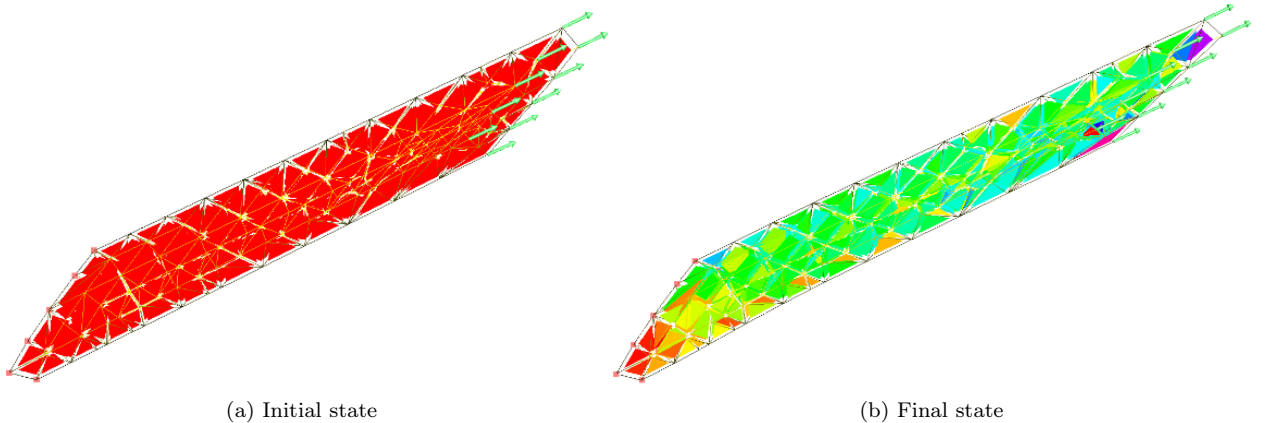


FIGURE 2 – The accessory band of dataset  $DS_2$  at two different simulation states.

All the simulations have been implemented in the Open Simulation Framework Architecture (SOFA)[2]; the complete datasets as well as the simulation scenes, example videos and the C++ code are available in the project's repository [3].

## 5 Results

Table 1 summarizes the obtained results. The five forearm datasets ( $DS_1$  to  $DS_5$ ) are listed with the corresponding ligament; CB stands for the Central Band, AB for Accessory Band, DOAC for the Dorsal Oblique Accessory Cord and DOB for the Distal Oblique Bundle. The second and third columns illustrate the values of the Young’s modulus  $\lambda$  and the yield forces  $f_y$ , respectively; values obtained from [1]. Value  $d_t$  is the measured displacement of the ligament in the tensile study, while  $d_s$  corresponds to the displacement obtained in the numerical simulations, which has been computed as the average of the nodal displacements of the nodes at the ulnar site. The last two columns correspond to the ratios between both displacements, for further comparison.

TABLE 1 – Displacement ratios – numerical simulations vs. reference study.

Dataset	$\lambda$ (Mpa)	$f_y$ (N)	$d_t$ (mm)	$d_s$ (mm)	$d_s/d_t$	$d_t/d_s$
$DS_1$ AB	25.52	156.196	4.68	20.55	4.39	0.23
$DS_1$ CB	6.44	68.8111	3.36	11.91	3.55	0.28
$DS_2$ AB	16.34	52.6312	2.42	4.47	1.84	0.54
$DS_2$ CB	5.03	132.604	7.01	23.14	3.30	0.30
$DS_2$ DOAC	16.93	38.8234	2.60	4.56	1.75	0.57
$DS_3$ AB	6.20	49.1	4.26	5.09	1.20	0.84
$DS_3$ CB	10.38	136.832	5.31	16.09	3.03	0.33
$DS_3$ DOAC	109.42	242.406	3.05	6.00	1.97	0.51
$DS_4$ AB	36.23	110.247	4.04	6.32	1.56	0.64
$DS_4$ CB	6.57	90.0987	6.14	16.21	2.64	0.38
$DS_4$ DOAC	46.35	74.042	3.29	4.78	1.45	0.69
$DS_5$ AB	16.74	84.754	4.41	7.59	1.72	0.58
$DS_5$ CB	12.35	210.688	3.84	16.00	4.16	0.24
$DS_5$ DOB	10.27	63.4192	4.64	9.44	2.03	0.49
Average					2.5	0.47

Further, the force-displacement curves have been plotted for each ligament with the forces ranging from 0 N to yield. The reference tensile experiment curve is plotted in blue against the simulation’s curve in orange. Results of the three diagrams for a sample Dataset ( $DS_2$ ) are illustrated on Figure 3.

## 6 Discussion

This study is an attempt to validate a numerical FEM simulation for individualized soft tissue behavior. The constitutive law is based on the Neo-Hookean approach, for which the morphological and biomechanical properties have been obtained from cadaveric experiments. The implemented simulation is very stable, robust, and free from volumetric locking. The simulation setup is simple enough to alleviate possible errors from boundary conditions, the numerical solver, or the collision handling. The latter may hinder the accuracy in case of complex collisions; it is not necessary in our setup due to the absence of contacts in the configuration.

Further, the reference tensile experiments and their setup have been performed with high fidelity; this includes the boundary conditions, the applied forces as well as the specific material properties.

Nevertheless, the obtained results indicate that the displacements which are obtained in our simulations are significantly higher than the tensile ones (on average 2.5 times larger). We believe that this could be due to one or a combination of the following reasons :

- Thickness of the ligament; the assumed regular thickness does not represent the morphology of the ligament in high fidelity. The ligamentous tissue is more dense at the regions of wrapping on the bones, and thinner in the central part. In order to investigate the impact of the thickness, we repeated the simulations for many ligaments with variations in thicknesses; we observed that this variation impacts the displacement directly. Note that the irregularity in the ligament’s thickness across its various sites was not taken into account in our models, and a regular thickness has been assigned for the ligament.
- The size of the volume elements in the ligaments’ meshes could slightly influence the simulation. i.e. when increasing the diameter of the sphere which includes a tetrahedron by 1.5 the displacement would be slightly increased (by about 10%). This suggests that smaller elements lead to reduced displacements. In general, in the volume meshing process we aimed at keeping the elements as small as possible.
- It could be that the constitutive law is not accurate enough for the given biomechanical experiment. Even though several other non-linear constitutive laws were technically available, such as Mooney Rivlin or Arruda–Boyce models, it was not possible to test them due to the absence of the required biomechanical parameters from the tensile tests; limiting the choices. Also note that the deformations which have been

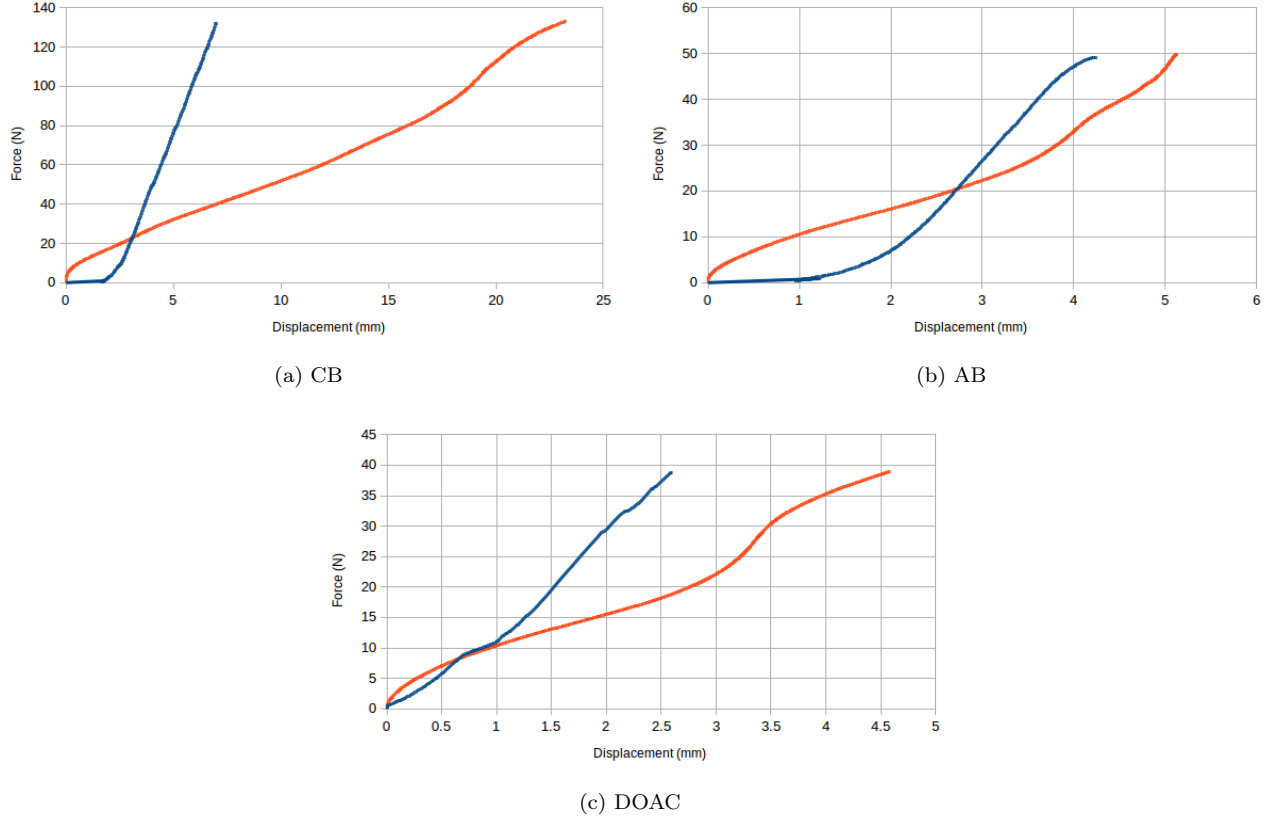


FIGURE 3 – Force-displacement curves corresponding to the ligaments in  $DS_2$  – tensile experiments in blue vs. numerical simulations in orange.

induced in the tensile tests are much higher than those which will occur in the normal forearm motion. Thus, at lower strain, Hooke’s law could approximate the linear part of the tensile experiment curve before reaching yield force.

- Another source of error could also be in the values of the Young’s modulus, the parameter which has the direct influence on the displacement order, which have been computed from the tensile setup and have been fed into the simulations. A variation from the average values reported in the literature has been observed, and the lower stiffness values have led to higher displacements in the simulations. It was possible in mostly all cases to reproduce the same displacement by selecting adjusted Young’s moduli.

To conclude, we think that the modeling of the ligaments should be improved to consider nodal contacts, resulting from wrapping ligaments around surface contact regions. Topological changes in the volume meshes could also be included, in order to reproduce the failure scenarios in the simulation. Further, more investigation should go in the direction of subject-specific parameter identification. Finally, the outlined discrepancies between experiment and simulation raised additional questions and concerns. Due to the complexity of ligament segmentation from patient imaging data, alternative modeling approaches may be required to achieve sufficiently accurate simulations. Moreover, it is not clear if relying on cadaver studies to validate the simulations will in the end be the best approach. It may be required to also consider in-vivo motion analysis with appropriate in-vivo imaging modalities.

## Références

- [1] Fabio Carrillo, Simon Suter, Fabio A. Casari, Reto Sutter and Ladislav Nagy, Jess G. Snedeker, and Philipp Fürnstahl. Digitalization of the IOM : A comprehensive cadaveric study for obtaining three-dimensional models and morphological properties of the forearm’s interosseous membrane. *arXiv e-prints*, 2020.
- [2] François Faure, Christian Duriez, Hervé Delingette, Jérémie Allard, Benjamin Gilles, Stéphanie Marchesseau, Hugo Talbot, Hadrien Courtecuisse, Guillaume Bousquet, Igor Peterlik, and Stéphane Cotin. SOFA : A Multi-Model Framework for Interactive Physical Simulation. In Yohan Payan, editor, *Soft Tissue Biomechanical Modeling for Computer Assisted Surgery*, volume 11 of *Studies in Mechanobiology, Tissue Engineering and Biomaterials*, pages 283–321. Springer, June 2012.

- [3] Noura Hamze, Fabio Carrillo, Philipp Fürnstahl, and Matthias Harders. Open dataset and simulations for soft-tissue tensile experiments. <https://zenodo.org/record/3728255.XoVHuvFS-93>, 2020.
- [4] Noura Hamze, Lukas Nocker, Nikolaus Rauch, Markus Walzthöni, Fabio Carrillo, Philipp Fürnstahl, and Matthias Harders. Automatic modelling of human musculoskeletal ligaments – framework overview and model quality evaluation. *arXiv e-prints*, 2020.