Finite Element Analysis of the Interaction between an Endo-Surgical Tool and Colorectal Tissue for Setting up Force Feedback Evaluation in Virtual Reality-Based Applications

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Abstract. Numerical simulations and Finite Element Analysis (FEA) have currently increased their applications in medical field for making preoperative plans to simulate the response of tissues and organs. Soft tissue simulations, such as colorectal simulations, can be adopted to understand the interaction between colon tissues and surrounding tissues, as well as the effects of instruments used in this kind of surgical procedures. This paper analyses through FEA the interaction between a surgical device and a colon tissue when it is fully clamped. Sensitivity analysis in the respect of the material mechanical behaviour, geometric approximation and the effect of thickness variation are investigated with the aim of setting up a virtual prototype of the surgical operation to aid mentoring and preliminary evaluation via haptic solutions. Through this investigation, the force feedback estimation that is necessary in many virtual-reality applications, may be estimated without discharging nonlinear effects that occur during clamping and that usually cannot be simulated efficiently to guarantee real-time solutions. Results are aligned with experimental data, confirming the reliability and right the set-up of FEA. Through them, the preliminary set-up of a haptic force feedback has been described and simulated through Simulink 3D animation, confirming the feasibility of the concept.

Keywords: Surgical simulation, FEA, Metamodeling, Virtual Prototyping, Force Feedback, Haptic Device.

1 Introduction

In recent years, virtual simulations have gained prominence in the medical field as a method for conducting preoperative planning and simulating tissues and organ responses. Many applications use Finite Element Analysis (FEA), especially to simulate real-world phenomena involving organs, surrounding tissues and surgical instruments (such as colorectal surgery simulations), as well as to design prosthesis, orthosis, analyse implants and simulations of surgery [1]. As a major drawback, FEA is not so efficient to achieve real-time simulation, as virtual reality-based applications ask for. Due to this, simplifications must be done to achieve interactivity, often reducing simulation's accuracy in terms of stress-strain distribution and force feedback. Furthermore, geometrical modelling and FEA may be principally used to set-up virtual investigations for the preoperative plan, on the base of the characteristics of the patient [2], but also, with the recent developments of additive technologies, to obtain printed 3D models useful for planning [3-5]. This kind of possibilities also gained importance in training and didactics applications, where learning by doing can be also associated to different levels of difficulties and accuracy. By this point of view, the linear elastic mechanical behaviour represents probably the most common assumption for simulating tissue deformation due to its low computational cost. Nevertheless, in soft tissues, well-known nonlinear effects are produced by hyper-elastic material behaviour and large displacements.

The aim of this paper is to investigate how nonlinear effects, evaluated through FEA, may be considered in a preliminary set-up of a virtual prototype of the colorectal surgery, where the response of the interaction between surgical tool and soft tissue is relevant. In section 2, a brief state of the art about simulation of soft tissues both in virtual reality and FEA environment is presented, highlighting open issues, then, in section 3, Materials and Methods, the investigated problem is presented. While, in section 4, Results and Discussion about their implementation for force feedback are presented and finally, in section 5, conclusions are outlined.

2 State of the Art

Surgical simulators based on virtual reality proves to be a good option to train new surgeons, to practice new processes, and to assist preoperative planning. For understanding tissues injury or critical physical behaviours during operations, accuracy of the mechanical mechanisms behind deformations is required. Many applications of virtual reality can be focused on the real phenomena that are present when there is an interaction between organs, surrounding tissues and surgical tools. In these pre-operative planning cases, stress accuracy is relevant similarly to the real-time feedback effect of the simulation, so that, equivalent mechanical models have been investigated [6].

The linear elastic mechanical behaviour is probably the most popular assumption to simulate the tissues deformation into virtual surgery simulators. It is based on Hooke`s law and has the advantage of needing only two constants to determine its behaviour (under the hypothesis of isotropic and homogenous materials), and low computational cost. These characteristics enables real-time haptic rendering. Unfortunately, real soft tissues behave hyperelastically. Many material model formulations are present in literature, as reported in [7], and they are often implemented in the most adopted FEA codes [8]. FEA commercial software cannot simulate in real time although they may provide result with better accuracy. In [9], a condensation method to extend a FE model for real-time surgical simulations is presented. This method suggests reducing the FEA model only at the set of nodes which undergo stressed so that evaluated displacement may be rendered. A comparison between the proposed method and the conventional linear FE analysis presents similar results in terms of nodal displacement. Another research [10] presents a real time simulation, thanks to a pre-processing of elementary deformations derived from a finite element method wherein the bulk of the computations were performed during the pre-processing stage of the FE calculation. This method was tested in a real-time hepatic (liver) surgery simulation.

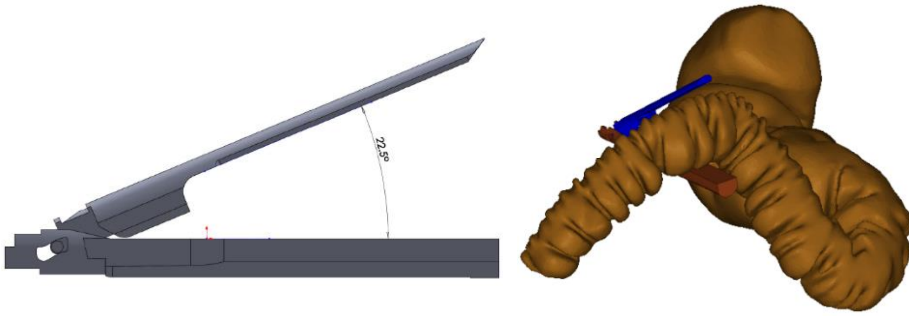
From the soft tissues point of view, better accuracy requests a nonlinear analysis that involves large displacement and a hyperelastic mechanical behaviour [11]. With the aim of a realistic behaviour in simulation for surgical training, in [12], researchers implemented the hyperelastic Mooney-Rivlin model. This work includes a proposal of a scheme for mesh adaptation based on an extension of the progressive mesh concept, so that real-time computation of deformation may be enabled.

Nowadays, it is possible to find software focused on the interactive computational medical simulations, as SOFA (Simulation Open Framework Architecture), which facilitates collaboration between specialists from various domains with the aim of approaching virtual reality [13]. To reach this goal, SOFA is organized in independent components in a scene graph data structure. Then, in each component it is possible to encapsulate a particular aspect regarding the simulation such as degrees of freedom (DOFs), forces and constraints, differential equations, main loop algorithms, linear solvers, collision detection algorithms, or interaction devices. In [14], SOFA is adopted to estimate in real-time, the position of the liver internal structures by evaluating nonlinear deformation of the liver, taken from images and vascular network simulated as vessels with simplified behaviour. Also in this case, the necessity of considering nonlinearity must be balanced with computational efforts.

From the virtual simulators point of view, it is possible to find several commercial solutions focused on surgical simulations. With the aim of a realistic simulation and a complete immersive experience, some surgical simulators provide a force feedback to the user by means of haptic systems. However, in many of these cases, the details of the adopted model for the soft tissues mechanical behaviour are not easily available. Linearization is often declared, tweaked on qualitative evaluation performed by few surgeons rather than actual material testing, highlighting the emphasis on producing models that are visually appealing. Recently, in [15], the problem of robot grasping during sutures is experimentally approached through haptic sensors suitable to assess the feeling at the finger. It investigates how visual and tactile information affects the evaluation of safety margin during tissue manipulations. In particular, two levels of indentation force are found assuming the presence or not of a visual feedback (respectively 75mN and 87.5 mN).

3 Material and Methods

Surgical clamping is a manipulation task necessary during a laparoscopic process. It does not involve tissue rupture but deformation able to hold in contact walls of the organ. In a laparoscopic process made in Mini-Invasive Surgery (MIS), instruments must be inserted through a trocar to access the abdominal cavity without capability of direct manipulation of tissues, so that, a loss of sensitivity occurs. This is caused by the absence of a direct force-feedback resulting from the interaction between surgical instruments and soft tissues in conjunction with an indirect field of view, as described in [16]. The force feedback received during the operation is a composition of the reaction force of the tissue and of the mechanical characteristics of the surgical tool. Fig. 1 shows a clamping tool in its opening position and the anatomical district of the large intestine and its FEA set-up when clamping is starting.



a) b)

Fig. 1. Test case: clamping device (a) and interaction (b) with the anatomic district (geometric model named as "3D-SS").

In Fig. 2a, an example of the kinematic chain is sketched. Green dot lines show the final position when the grasp is close. Cam geometry, sketched in the detail, and spring stiffness (named as *k*) characterize the quality of the tool. Fig. 2b describes the dynamic equilibrium, so that the force at the latch constrained in *A*, F_i, may be found as a function of the tissue reaction (F_g).

The clamping device may be assumed as rigid considering that, assuming a maximum allowable pressure of $8*10^{-2}\ N/mm^2$ at the interface with the tissue [17], FEA results on the final part of the clamping device show a maximum deflection of about 0.4 mm. According to this, the relation among the momentum M_a induced by F_iL_3 is related to the F_g at the interface with the tissue by:

$$F_g = F_t'\ m/n \tag{1}$$

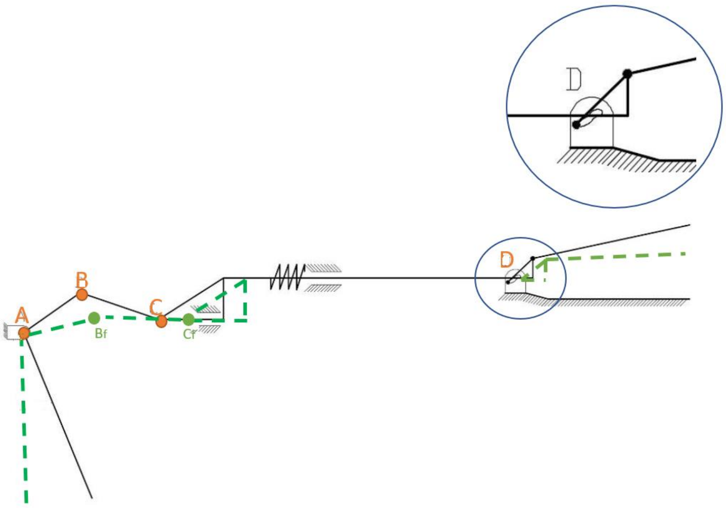
where F_t' stands for:

$$F_t' = \frac{F_i * L_3}{L_1} * \cos\varphi * \cos\beta - k(\Delta x) \tag{2}$$

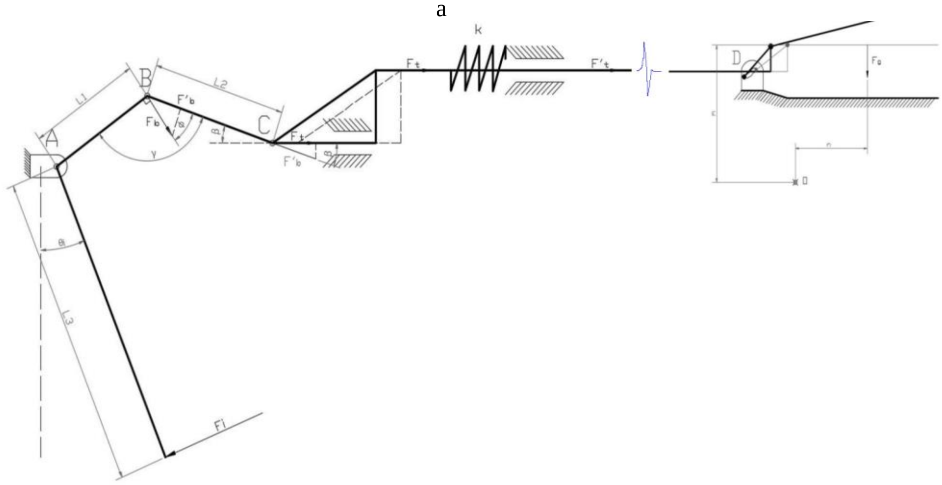
where Δx represents the relative displacement of $C(\theta_i)$ from $C_0(\theta_i=22°)$ and φ is function of θ_i that spans the range [0 to $\theta_i/2$]

$$\sin\beta = \frac{\left(d\cos\theta + \sin\theta\sqrt{L_1^2 - d^2}\right) - d}{L_2} \tag{3}$$

$$\varphi = \frac{\theta_i}{2} \tag{4}$$



a



b

Fig. 2. a) Kinematic chain, b) reaction force scheme.

The interaction with the soft tissue was investigated through FEA, applied on three different geometrical models, as discussed in [8]. The most accurate is the one shown in Fig. 1b, named 3D-SS, while an approximation of the real geometrical model is named 3D-S and finally the simplest is the cylindrical approximation of the region of interest, named CS, with diameter equivalent to the average one of the final segments.

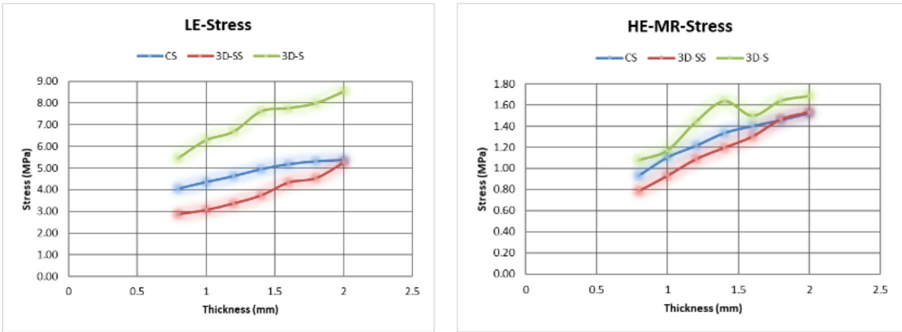
FEA has been carried out in Altair HyperWorks suite using RADIOSS solver. RADIOSS have been chosen due to its advantages in handling contact algorithms and large displacements, since it is an explicit solver oriented to minimize computation time in FEA with high non-linearities [18]. Other parameters and input conditions have been set as described in the following:

- Mesh: Shell elements with three integration points along the thickness (named shell3N and shell4N) have been used to construct 3D-SS, 3D-S, and CS geometrical models.
- Load & Boundary conditions: it has been assumed the usage of a blue staple cartridge with a 1.5 mm closed height [19]. This gap has been adopted as final closure of the tissue under the clamping device, assuming no sliding at the interface. Concerning displacement constraints at the free edges of the models, rotations around all axes, as well as the displacement along the colon axial direction at the nodes along the free edges have been released, in order to reproduce constraints, close to those present in tissue when conducting colorectal surgery.
- Contact: The "Type 7 interface" into HyperWorks is the contact condition defined between the colon tissues and the surgical clamp, which is a multi-usage impact interface. To ensure that the simulation developed correctly and to establish self-contact and contact between colon tissues and the surgical clamp in the current case study, more than one "Type 7 interface" was used.
- Soft tissue's thickness: it is related on the healthiness of tissue. The colon wall thickness value is set in the range of 0.8 to 2 mm to consider all assumptions above mentioned, these values are based on data available in literature [20,21].

To highlight the role of the material model in the stress-strain behaviour, thus in the force feedback, both Mooney-Rivlin [22] (HE-MR) and linear elastic (LE) material models have been considered in this paper. In [2], the Constant values are set to C_{10} = 0.085 MPa and C_{01} = 0.0565 MPa. These values are sufficient to define the hyperelastic Mooney-Rivlin model when the used materials are incompressible. In the same way, several applications require a reasonable approximation of soft tissue mechanical properties. For this reason, some laboratory tests [23] have found an experimental approximation to the linear mechanical behaviour corresponding to E = 5.18 MPa.

4 Results and Discussion

Fig. 3 highlights the comparison in terms of stress by changing geometrical model and tissue thickness, in the respect of LE and HE-MR material models.

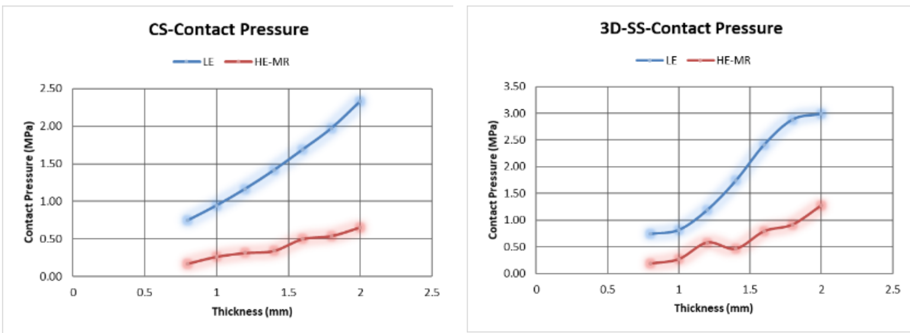


a) b)

Fig. 3. Trends of stress varying thickness and geometrical model for linear elastic material model (on the left), and for hyperelastic MR one (on the right).

The first consideration concerns the range of stress passing from LE to HE-MR. LE overestimates the maximum stress under the clamping device, despite the adopted geometrical model. The second one concerns with the irregularities of the 3D-S model in the respect of the others, considering that in many cases CS tends towards 3D-SS. The differences between the stress ranges are related to a major capability of stress distribution along the areas of the HE-MR model in the respect of LE. It means that the stress gradients are more gradual in the HE-MR case, as already discussed in [11]. Thickness increase always induces a stress increase.

Fig. 4 compares the most stable solutions, CS and 3D-SS, in terms of contact pressure with the grasp, changing material model, thickness and geometrical model. Contact pressure increases with the changes of the initial thickness. The average values in the case of 3D-SS and CS models, assuming HE-MR is about 0.5 MPa, with a trend that can be reasonably seen as linear in the respect of the initial thickness changes. The LE material model, on the contrary induces a major stiffness, thus higher contact pressure values.



a) b)

Fig. 4. Trends of contact pressure varying thickness and material model: a) CS geometrical model and b) 3D-SS geometrical model.

It has to be noted that the contact pressure values are referred to a limited area of elements, taking the maximum values between those instantly in contact with the clamp. This fact is understandable looking also to the pressure force trend (Fig. 5).

FEA analysis made through Radioss follows the dynamics of the contact explicitly allowing the computation of force and displacements along time, so that, Force-Displacement curves may be derived changing geometrical models, materials and thickness. Fig. 5 shows the contact force during clamping as a function of the displacement of the clamping tool, up to put in contact the upper and lower part of the colon section. This induces saturation of the contact force, after a plateau of less than 1 N (in the blue squared detail). This graph describes the reaction force on the grasp that the tissue applies under assigned displacements. It is consistent with the results found in [16] and the contact pressure found in the previous section.

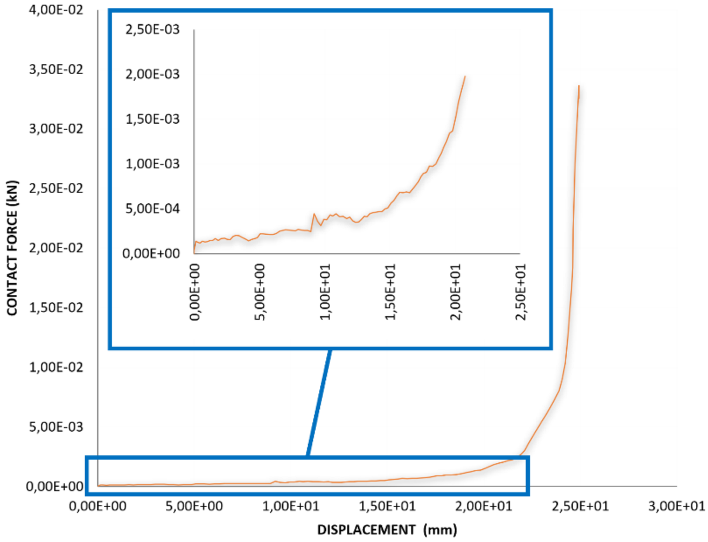


Fig. 5. Contact force vs vertical displacement, and, in the blue squared detail, a zoom of the plateau part of the curve.

Table 1. The kinematic chain's starting values.

Identification	Magnitude
L_1	21 mm
L_2	25.1 mm
L_3	54.1 mm
$\theta_{initial}$	22º
$\beta_{initial}$	19º
φ	$\theta_i/2$

Through the dynamic balance described in Section 3, the reaction force at the hand interface can be computed in accordance with the linkage lengths as reported in Fig. 2 and defined in Table 1.

Fig. 6 shows the percentage values of F_g in the respect of F_i. This value has been adopted for a preliminary implementation of a virtual environment able to provide the force feedback during clamping. It has been designed in Matlab through Simulink 3D Animation and it will be interfaced with an Arduino's controller able to provide the feedback to a simplified clamping device achieved by additive manufacturing. Fig. 7 shows the concept diagram of the system, where the force feedback function is based on the input provided from a post processing of the FEA collected in this work, starting from the CS model.

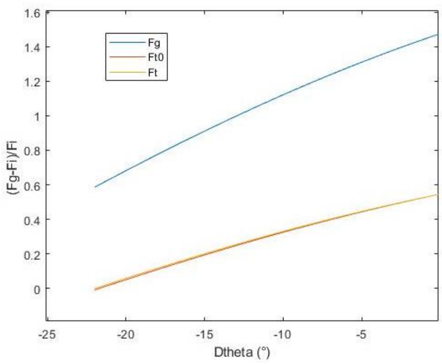


Fig. 6. Angular displacement (θ) vs F_g.

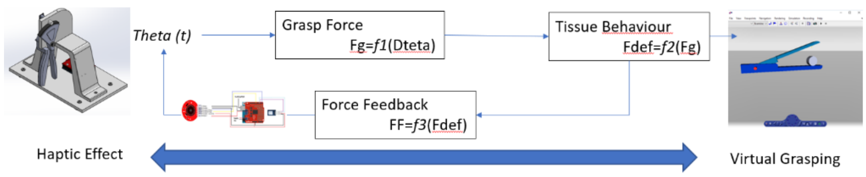


Fig. 7 Concept blocks of the proposed haptic device.

More in details, the VR sink implements a collision detection and the CS deformation derived from data collected in Fig. 5. The Simscape model of the grasping tool (Fig. 8) has been simplified having in mind the chance of implementing a haptic sensor through pulley, able to implement the force feedback on the base of an open loop impedance controller. Fig. 9 captures the beginning of the contact and two simulation states along the contact force curve, assuming a half model of the CS so that the visual correlation among theta and grasp displacement may be seen.

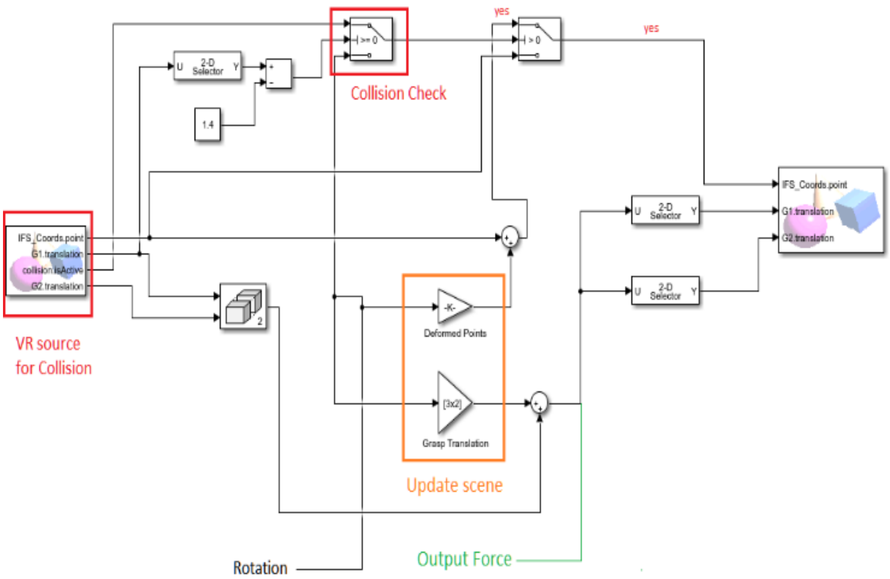


Fig. 8 Virtual Grasping modelling via Simulink 3D Animation**.**

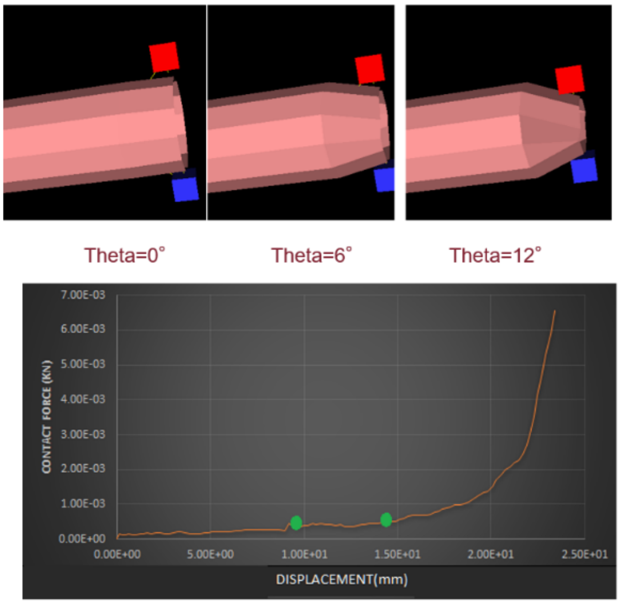


Fig. 9 VR results of the correlation among Theta-Displacements

5 Conclusions and Future Works

This paper presents the stress-strain analyses of the surgical grasping of soft tissues in correlation to the force feedback that should be provided in haptic solution for virtual surgical experience. More in details, starting from previous analyses made about the role of hyperelasticity, accuracy of the geometric models and tissue initial thickness, FEA model based on Mooney-Rivlin hyperelasticity have been proved to achieve results aligned with values found in literatures. They confirmed a rather linear behaviour in the adopted range of simulations. Through these results, a haptic device with an open loop impedance feedback has been designed and preliminary presented. In particular, starting from a reasonable configuration of the grasping device, the force at the handling has been evaluated as a function of the imposed rotation and the related reaction force under the tissue has been estimated by the FEA. Then, concept blocks and test simulations through Simscape-Simulink animation 3D have been built. They confirmed the feasibility of the solution, starting from the CS geometrical model and the linearization made from the FEA results. The final goal is providing a VR environment able to support surgical testing through material behaviour information related to stress and strain and force feedback. Next developments will include the construction of the hardware and the testing of the controller that will be controlled through Arduino.

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