Interaction modeling and simulation of a flexible needle insertion into soft tissues

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Abstract

Needle puncture is irreplaceable for some percutaneous therapies, like biopsy, brachytherapy treatment, neurosurgery, etc. But needle insertion force deforms soft tissue, making accurate targeting difficult. It thus emerges as an important issue to model the interaction force raised during needle insertion. In this paper, we establish the force model with data acquired from the simulation of a flexible needle insertion into a liver of human being. First, high-quality tetrahedral meshes of the flexible needle and the liver are generated using an adaptive constrained *Delaunay* tetrahedralization algorithm. Then, the tetrahedral mesh data is integrated into SOFA, an open-source framework, to dynamically simulate the interactive behaviours of the flexible needle insertion into the liver. Finally, interactive force data acquired from the simulations is identified and modelled in three parts, namely, stiffness, friction, and cutting forces. The models established may be used in robotic needle steering and insertion training with computer simulator.

KEYWORDS: flexible needle, soft tissue, interaction modelling, FEM simulation

1 Introduction

Needle puncture has already become a substantial branch of medical procedures around the world, not just in China. Percutaneous therapies using needles are appropriate for medical diagnosis and treatment. Needle lumen provides a conduit through which to deliver drugs, radioactive seeds, and collect samples from deep-seated tissues. But, manual needle insertion relies on the physician's kinaesthetic feedback, and inevitably results in targeting error, and requires multiple insertions, which cause significant outcomes, like excessive healthy tissue trauma. Robotic needle steering appears to be promising to improve targeting accuracy. It overcomes the limitation of human errors due to fatigue, hand tremor. Furthermore, it enables physicians to be away from X-ray exposure. In the last ten years, robotic needle insertion has received much attention, and finds potential applications in biopsy, brachytherapy, cryogenic ablation, electrolytic ablation, deep brain stimulation and minimal invasive surgeries, etc.

Robotic needle insertion improves substantially targeting accuracy. Recently, needle insertion robots for prostate biopsy and breast tumor ablation have developed in laboratories. However, substantial effort is needed to reduce targeting error and to make these robots clinically viable. Targeting displacement is always crucial for needle puncture, and it is the main cause of constraining these robots' applications. Visual limitation, physiological movement, tissue deformation and needle deflection result in gross targeting errors in a percutaneous insertion. Traditionally, intervention needles are rigid, and penetrate tissues in a straight-line path, which makes it difficult to avoid obstacles in the path, like bone, vessel, nerve, and other vital organs, as needle misplacement happens. A feasible solution to mitigate the misplacement is to use flexible needle that can be robotically steered to deviate from the penetrating direction, and reach the specific target that is inaccessible to rigid needle due to anatomical obstacles. The steerability of a flexible needle is attributed to the asymmetry of the bevel tip, which produces bending forces at the needle tip, and the flexibility of the long and slender needle shaft, which allows the needle to follow a curved path when pushed forward. The tissue, into which the flexible needle is inserted, also contributes to needle bending. So, through spinning the flexible needle at its base, the bending forces are redirected, and result in the orientation deviation at the needle tip from the straight line path. However, it is very challenging for a robot to precisely control the targeting deviation of a flexible needle.

Medical simulator is of great advantage of training medical students, who are required to practise needle insertion techniques in a controlled environment. Usually, a large number of attempts are conducted on patients, mannequins for a fresh hand to become proficient. This computer-aided simulator can help the training and avoid incidents, even rehearse a complex insertion and control a medical robotic system. The success of an insertion simulation depends largely on an accurate model. Therefore, in this study, we focus on simulating a flexible needle insertion into soft tissue, and modelling needle-tissue interactive forces.

2 Related work

Needle insertion simulators comprise of soft tissue model, needle model and needle-tissue interaction model. Over decades, lots of work on needle insertion modelling has

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been published, but, separate models are generally used, that is, interactive forces, tissue deformations and needle flexibility have not been considered together. Earlier works were present by Okamura et al. [1-2] and Altervitz et al. [3]. Okamura et al. identified needle forces into stiffness force before tissue capsule is penetrated, and friction force and cutting force after a capsule penetration, from the experimental data. Webster et al. [4] presented a non-holonomic kinematic model for a flexible needle with bevel tip. A mechanics-based model of robotic needle steering is established by Misra et al. [5, 19], based on macroscopic and microscopic force observations at the needle tip. However, these methods do not considered the complex boundary conditions that are normally present. Recently, needle-tissue interaction has been studied based on mass-spring simplification [6-7] and finite element method (FEM) [8-12].

FEM is the most common technique to model deformable tissues. Hing et al. [8] used the displacements of implanted fiducials inside soft tissue and the needle insertion force to calculate interactive parameters, which characterize force versus position profile of a needle insertion and withdraw task through a reality-based modelling with a finite element tissue model. DiMiao et al. [9] simulated flexible needle insertion in 2D space using FEM with meshes of triangular elements, and in order to reduce the computational complexity and overcome the curse of dimensionality, a condensation technique was proposed by ignoring the nodes that are far away from the needle shaft. This method was extended to 3D space using tetrahedral elements by Goksel et al., [10]. In [11], they further presented three models to simulate the deformations of a flexible needle. The first two models use FEM with tetrahedral elements and nonlinear beam elements, respectively. The third is modelled with rigid bars connected by angular springs. Comparison was made to the simulation results, but find that the angular spring model is computationally efficient, and the most accurate to compute the deformations of the needle. Differ from other medical tool-tissue interactions, the coupling underneath organ surface makes it difficult to measure and simulate the forces along the needle shaft as it inserts into soft tissue. To this end, Chentanez et al. [12] introduced new algorithms for simulating and visualizing the needle insertion and steering in deformable tissues through local remeshing, and coupling 3D finite element tissue model and 1D finite element needle model with stick-slip friction. Remeshing and coupling ensure that the contact nodes dynamically have same position in the meshes of soft tissue and needle.

But, remeshing is theoretically complicated and computationally expensive, especially for high order elements. Mesh modification, on the other hand, use local adjustment of existing nodes to approximate the needle's motion, which reduce the computational cost but sacrifice accuracy. Online mesh modification is more challenging. Therefore, it is difficult to realistically simulate a needle insertion at fast frame rates. Xu et al. [13] proposed a meshless approach. Without using finite elements, two sets of unstructured nodes are used to represent needle and soft tissue. Mesh modification and remeshing were therefore not needed and fast simulation was realized instead. In [14], a generic method based on dedicated complementarity constraint models was proposed to simulate the insertion of a flexible device into soft tissue without remeshing, in 3D space. Granted, we herein use the surgical simulation toolkit SOFA [17] to simulate a flexible needle insertion to soft tissue, and establish the interactive model from the force data acquired during the simulation.

3 Tetrahedral Meshes of Soft Tissue and Flexible Needle

In this simulation, FEM is used to numerically solve the partial differential equations (PDEs) of a flexible needle insertion into soft tissue (liver from human being is used here). Its accuracy depends on the element discretization geometry. Simultaneously, the speed for solving PDEs is determined by the number of the discretized elements as well. It is therefore important to discretize the liver and the needle into appropriate geometric meshes, and high-quality meshes are essential to accurately and efficiently simulate an insertion procedure. The following steps are taken to produce these meshes.

First, the geometric mesh models of the live and the flexible needle are constructed respectively, as shown in Fig.1. The geometric model of the flexible needle is obtained using the 3D animation suite of *Blender*. Alternatively, considering the anatomic complexity of a liver, its geometric model is directly taken from SOFA, which is an open source framework for real-time simulation, with an emphasis on medical simulation. SOFA has built numbers of complex organ models, and these models are generated from 3D biomedical imaging data.

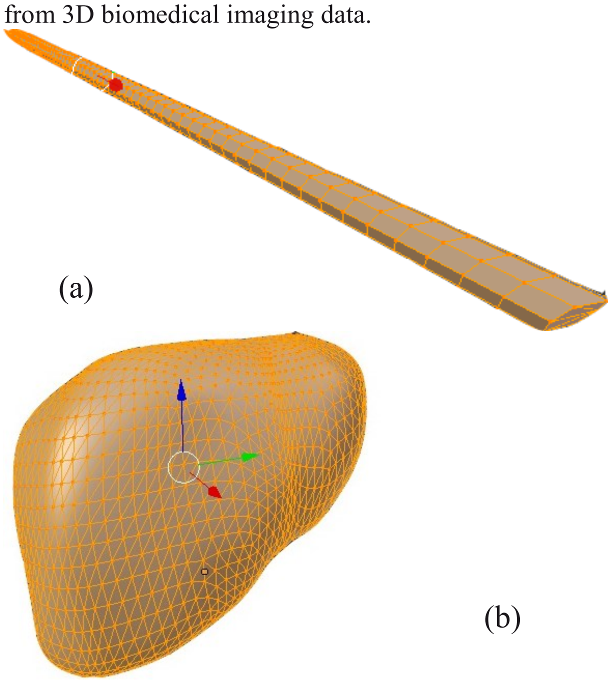

Fig. 1 Geometric models data: (a) the flexible needle, (b) the liver.

Second, generate meshes of tetrahedrons with the geometric models data. To guarantee accurate simulation result, these tetrahedral meshes must be well shaped, and their size and quantity should be under control. *Delaunay* tri-

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angulation [15] is widely used to generate tetrahedral meshes of geometry volumes, and this technique is effective both in theory and in practice, and is very successful in 2D mesh generation. But, a notable problem for the traditional *Delaunay* algorithm is that the generated tetrahedrons may contain badly shaped tetrahedrons, namely, the so-called silvers, which are flat. Silvers have 0^{o} or 180^{o} dihedral angle, and their volumes are close to zero so that arbitrarily big simulation errors may be aroused. Therefore, mesh optimization and smoothing is required to remove these silvers and improve the overall mesh quality. As introduced above, it is desirable to generate smaller tetrahedrons, but computational cost is dramatically increased. Adaptive method was thus introduced to find a balance between mesh size and computational cost [16]. It inserts new points such that the constructed mesh is well shaped, and its size conforms to a specific sizing function. In [16], the constrained *Delaunay* tetrahedralization (CDT) is proposed to optimize and refine meshes, and certainly to remove silvers. TetGen, based on CDT, is used here to generate tetrahedral meshes for both the flexible needle and the liver in the spatial geometry domain, as shown in Fig. 2.

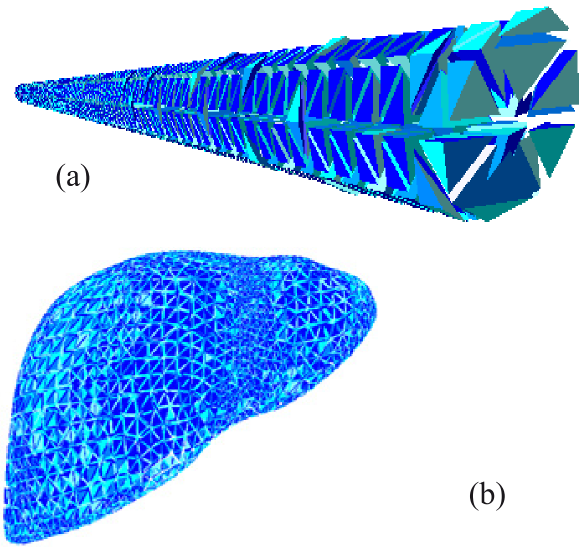

Fig.2 The tetrahedral meshes: (a) the needle, (b) the liver.

4 Flexible Needle Insertion Simulation with SOFA

Flexible needle insertion simulation is implemented in this study using SOFA. It is a modular framework applied to surgical simulation, and consists of visual, collision and behaviour models. The visual and behaviour models receive data of the tetrahedral meshes generated in the previous section. But the visual model is used for visualization only. Each model can be designed independently of the others, but their consistency is maintained using mapping. Mapping enables us to use different geometric models for a specific object.

Herein, SOFA is used to simulate the interactive behaviour of a flexible needle insertion into soft tissue, and acquire the interaction force data. Figure 3 shows the scene graph composed of a flexible needle and a liver. It is a hierarchical structure. The liver and the flexible needle nodes under Root have children nodes and components. Each has three models mentioned above. The mechanical object, namely the behaviour model, is the most important component, which contains information of coordinates, velocities, forces and acceleration. It communicates with other two models and exchanges parameters data. Contact force is generated as the flexible needle penetrates into the liver. To acquire the interactive forces data, collision model is required to create *a priori* to detect collision or contact between the needle and the liver. There are several collision models provided by SOFA, for example, sphere, triangle, line and point model. Considering that the flexible needle inserts into the interior of the liver, and sometimes no collision happens on the mesh edges in point model, line collision model is employed in this simulation, to ensure collision detection at these edges.

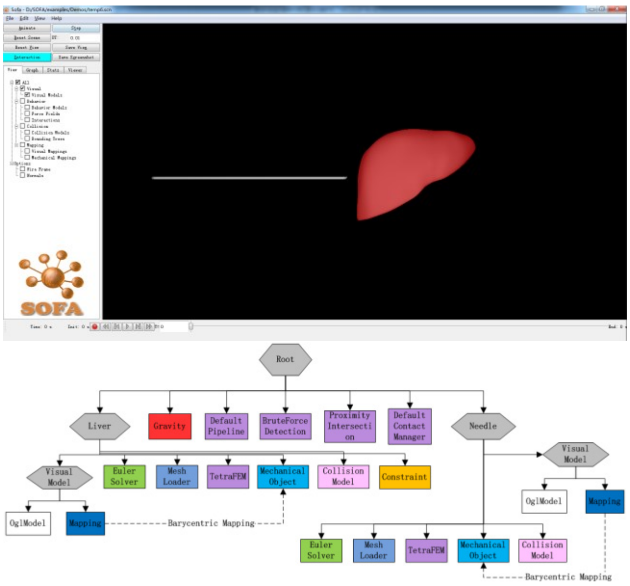

Fig.3 The scene graph of a needle insertion into soft tissue.

To complete the interactive simulation, Euler solver is applied here. It is a simplest, fast method to solve the finite element PDEs with satisfied frame ratio. An important parameter in collision model to decide the convergence of the solver is contact stiffness (CF). CF corresponds to material hardness. The liver deformation and the flexible needle bending are shown with various CFs in Fig.4. It can be seen that bigger CF constant results in larger liver deformation, and bigger interactive forces.

5 Interaction Modelling of Flexible Needle Insertion into Soft Tissue

In the recent years, robotic needle steering has aroused a surge of interest in the community of robotics, and needle-tissue interaction modelling has been investigated intensively with data from experiments, for use in the control of steering robots [18]. Such models are useful for realistic surgical simulations, too. But, it is very difficult to collect experimental data from *in vivo* organs due to medical risk. Moreover, no tactile/force sensor is availa-

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ble to directly measure the interactive forces on instruments in clinic. Currently, realistic data are typically collected from *ex vivo* tissues or phantoms. Contrast to these reality-based models, the data of this study come from the needle insertion introduced previously, and are used to model the needle-tissue interaction. Although simulation is conducted on the basis of a compromise of material fidelity and computational efficiency, it is still able to establish the force-displacement profile of the flexible needle and determine the way soft tissue deforms during contact with the flexible needle.

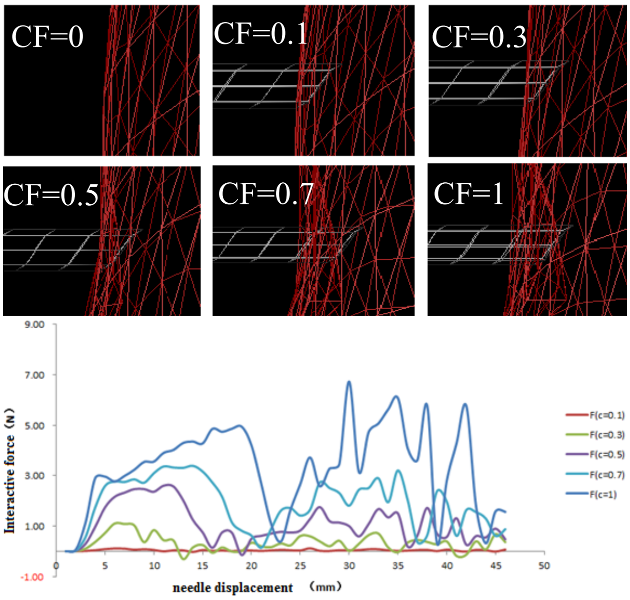

Fig.4 Liver deformations and interactive forces with various contact stiffness.

Liver is used here as the organ of interest, through which a flexible needle is inserted. The insertion progress in the simulation may be divided into three phases. At the beginning, the flexible needle deforms the liver surface as pushed forwards, without penetrating the liver capsule. When the force applied to the liver reaches a marginal threshold, the flexible needle is penetrated into the liver interior, and starts cutting the liver tissue. Finally, the flexible needle is required to completely penetrate through the liver. Therefore, the interaction forces can be analysed accordingly in terms of the three needle insertion phases [2]. As shown in Fig.5, in the first phase, a main puncture can be identified by a peak in force after a steady rise, followed by a sharp decrease. Therefore, stiffness force is only generated during this phase. The interactive force of the second phase results from the cutting and friction forces. The cutting force is to rupture interior tissue of the liver, while the friction force occurs along the needle shaft due to tissue adhesion and damping. Apparently, in the third phase, only friction force is left, without stiffness and cutting forces.

These interactive forces are modelled independently. First, the stiffness force is expressed by a nonlinear polynomial, since the liver capsule and its interior tissue are typically viscoelastic. It has the form of

$$F_s(x)=a_1x+a_2x^2 \qquad x<d_1 \qquad (1)$$

where $F_s(.)$ represents the stiffness force, x is the needle displacement, and d_1 is the maximum displacement of the needle before the liver capsule is penetrated. a_1 and a_2 are constants to be determined.

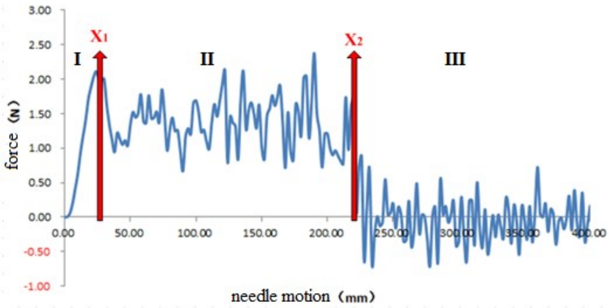

Fig.5 interactive forces profile of a flexible needle insertion into soft tissue.

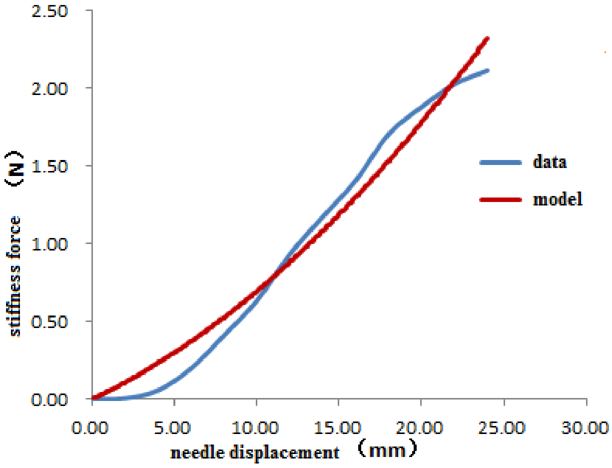

Fig.6 the fitted stiffness force and data acquired from simulation.

Using the force data before penetration, the second-order polynomial is fitted, shown in Fig6. It can be seen that the interactive force in the first phase can be closely approximated with the polynomial. Its constants of a_1 and a_2 are 0.0019 and 0.0499, respectively.

During the second phase of the flexible needle insertion, the interactive force comprises of cutting and friction forces, but, it is really difficult to isolate them. So, we model here the friction force of the third phase in advance, and postulate that it is same as that of the second phase. Friction is often defined with the Karnopp model. In [2], a modified piecewise Karnopp friction model was presented to describe the prominent frictional forces during needle insertion. But, the relative movement between the needle and the liver interior tissue is often invisible. Therefore, it is very difficult to estimate the parameters of the friction model.

Coming back to Fig.5, we can see that the friction force in Phase III displays some vibratory property of wave. In essence, it is reasonable since the liver tissue is viscoelastic and inhomogeneous. Here, a Fourier series form is then used to model the friction force with the first several dominant frequencies, given by

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$$F_f(x) = \sum_{i=0}^{M}[A_i con(\omega_i x) + B_i \sin(\omega_i x)] \qquad x>d_2 \qquad (2)$$

where A_i and B_i are Fourier coefficients, which are determined from the data of the needle insertion, ω_i is the truncated frequencies of vibratory friction force, M is the order of the Fourier series, namely, the number of the truncated frequencies, and d_2 is the needle displacement to penetrate through the liver tissue.

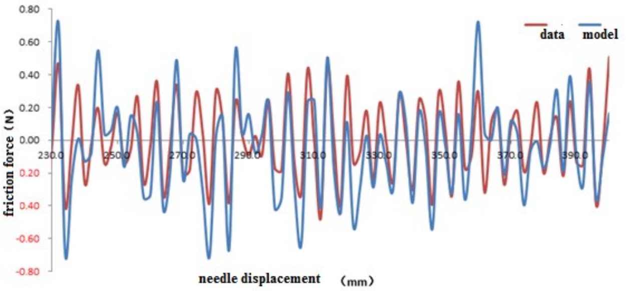

Fig. 7 the modelled friction force and the data of the third phase in the simulation.

The Fourier series form is superior over other fitting methods to approximate the experimental or simulation data. It could be an alternative to the previous polynomial. Figure 7 shows data acquired in the third phase of needle insertion simulation and the result of modelling friction force. Fourier transform is used to analyse the frequency spectrum of the obtained force data, and the first two dominant frequencies are truncated here to model the friction forces after capsule penetration.

The cutting force can therefore be determined by subtracting the friction force from the force data acquired in the second phase of needle insertion, and is given by

$$F_c(x) = \begin{cases} 0 & x \le d_1 \; and \; x \ge d_2 \\ F_a(x) - F_f(x) & d_1 < x < d_2 \end{cases} \qquad (3)$$

where $F_c(.)$ is the cutting force, and $F_a(.)$ is the measured value of interactive force.

Now, a complete model of the insertion force is able to be established, and is compared to the simulation data, as shown in Fig. 8. Although the profile of the model is nearly close to the data, it is noted here, the variations of the liver geometry and interior structure, the flexible needle deflection, the insertion speed, etc., could result in remarkable modelling errors. But, the established model can still be applied to the simulator for needle insertion training, and robotic needle steering.

6 Conclusion and Future Work

In this paper, we study the dynamics behaviours of a flexible needle insertion into a human being liver, and establish the interactive force model with data acquired from the simulation. First, to remove flat tetrahedons, a constrained *Delaunay* tetrahedralization algorithm is used to generate high-quality tetrahedral meshes of the flexible needle and the liver, respectively. Then, the tetrahedral meshes are integrated into SOFA to dynamically simulate the flexible needle insertion into the liver. Finally, interaction force data collected from the simulations, is modelled with stiffness, friction, and cutting forces. The stiffness force occurs before the liver capsule is penetrated, and is expressed with a polynomial, which deforms the liver tissue. After the capsule of the liver is penetrated by the flexible needle, the friction force along the needle shaft occurs, and is defined with Fourier series. The classical Karnopp friction model is not used here due to the difficulty to measure the relative velocity of the needle and the interior tissue of the liver. The cutting force is to ruputure the liver tissue, into which the needle is inserting, and is determined by subtracting the friction force from the gross froce.

The models, established from the simualtion data, will be used in the control of a needle steering robot, and be experimentally verified through *ex vivo* tissues and phantoms. A test rig of robotic needle steering system was already constructed, and will be used in future work.

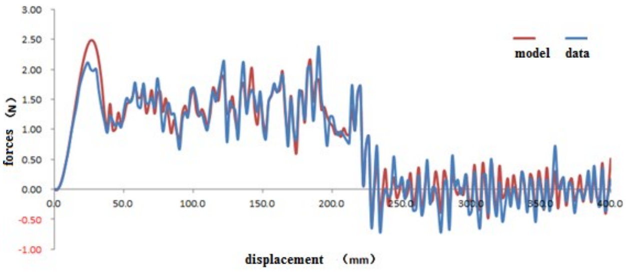

Fig. 8 the complete needle insertion model is compared to the simulation data.

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