The Nonlinear Material Properties of Liver Tissue Determined
From No-Slip Uniaxial Compression Experiments

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Abstract
The mechanical response of soft tissue is commonly characterized from unconfined uniaxial compression experiments on cylindrical samples. However, friction between the sample and the compression platens is inevitable and hard to quantify. One alternative is to adhere the sample to the platens, which leads to a known no-slip boundary condition, but the resulting non-uniform state of stress in the sample makes it difficult to determine its material parameters. This paper presents an approach to extract the nonlinear material properties of soft tissue (such as liver) directly from no-slip experiments using a set of computationally determined correction factors. We assume that liver tissue is an isotropic, incompressible hyperelastic material characterized by the exponential form of strain energy function. The proposed approach is applied to data from experiments on bovine liver tissue. Results show that the apparent material properties, i.e., those determined from no-slip experiments ignoring the no-slip conditions, can differ from the true material properties by as much as 50% for the exponential material model. The proposed correction approach allows one to determine the true material parameters directly from no-slip experiments and can be easily extended to other forms of hyperelastic material models.

Keywords: unconfined compression; soft tissue; liver; mechanical properties; friction
I. INTRODUCTION

The mechanical characterization of liver tissue can be highly beneficial in the management of liver diseases and injuries. In the U.S. alone, more than 20,300 people die each year as a result of complications arising from chronic liver diseases and the associated fibrosis and cirrhosis\(^1\). Also, the liver is the most frequently injured abdominal organ in motor vehicle accidents and falls from heights, both of which result in rapid decelerations\(^2\).

Unconfined uniaxial compression, a schematic for which is shown in Fig. 1, is a prevalent method for characterizing the nonlinear response of liver\(^3,4\), kidney\(^4,5\), brain\(^6\) and other types of soft cellular tissue because these types of tissue lack significant tensile stiffness. Also, the state of stress induced in uniaxial compression is simpler than indentation\(^7\) and aspiration\(^8\) experiments, and an exact solution exists for the underlying mechanics problem provided there is no friction between the compression platens and the tissue sample. In practice, however, friction is inevitable, not easily measurable, and must be accounted for in order to accurately characterize the material.

![Schematic of the uniaxial compression experiment.](image)

Of course, the issue of friction in these experiments is not new and has received attention from many researchers in the last two decades. Armstrong et al.\(^9\) conducted unconfined compression experiments on articular cartilage and concluded that the frictionless condition is “practically impossible” to achieve. Spilker et al.\(^10\) used finite element (FE) simulations of unconfined compression to show that the effects of friction can be significant, particularly as the specimen’s diameter-to-height ratio increases. Wu et al.\(^11\) arrived at a similar conclusion.

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Miller\textsuperscript{12} showed that even a coefficient of friction of 0.1 results in non-negligible increases in reaction forces for a material similar to brain tissue. Mendis et al.\textsuperscript{13} also considered the response of brain tissue and observed significant stiffening of the stress-strain response in their simulations in the presence of a no-slip boundary condition between the specimen and the compression platen.

To take the effects of friction into account, Wu et al.\textsuperscript{14} proposed a procedure in which the coefficient of friction is experimentally determined using custom apparatus and then utilized in finite element analyses to extract the real material parameters. In contrast, Miller\textsuperscript{12} suggested bonding the test sample to the compression platens to create a known no-slip boundary condition (see Fig. 1), and then correcting the measured stretch ratio to account for the no-slip condition. This correction is derived analytically, assuming that planes normal to the direction of the applied force remain plane during compression. For linearly elastic isotropic materials, Chau\textsuperscript{15} derived a correction factor for the Young’s modulus determined from no-slip experiments assuming the Poisson’s ratio is known. However, Chau’s result is of limited use for the problem at hand, given the highly nonlinear nature of soft tissue.

This paper adopts an approach similar to Miller’s in that we propose adhering the sample to the compression platens and then accounting for this through a set of correction factors, one for each material parameter. The correction factors are derived computationally and therefore it is not necessary to make assumptions regarding the state of deformation in the compressed sample. The correction factors allow one to determine the true material properties of soft tissue directly from no-slip uniaxial compression experiments.

Based on extensive finite element analyses, we derive relationships between the apparent mechanical properties of liver tissue (found from no-slip tests) and its actual mechanical properties. For this purpose, we assume that bovine liver is incompressible, isotropic, hyperelastic and that it is characterized by the exponential form of strain energy density function. Though we only consider one particular type of hyperelastic material model here, the approach itself is quite general and can be used with a wide variety of hyperelastic material models.

We apply the proposed correction factor method to data from no-slip uniaxial compression tests on bovine liver. The “corrected” stress-strain response is compared to that from (nearly) frictionless uniaxial compression experiments. We also test this approach by applying it to uniaxial compression data from Chui et al.\textsuperscript{16}, where the no-slip boundary condition
was not taken into account in characterizing porcine liver tissue. In addition, we compare the results from our correction approach to those from Miller’s correction\textsuperscript{12}.

II. A “CORRECTION FACTOR” APPROACH FOR CHARACTERIZING TISSUE

A. Continuum Mechanics

Tissue from most cellular abdominal organs, such as the kidney and the liver, is considered to be isotropic, (visco-)hyperelastic, and incompressible\textsuperscript{17,18}. The mechanical response of such materials is best described within the continuum mechanics framework, using a strain energy function, $W$, from which a material constitutive behavior is derived. When the viscoelastic and temperature related effects are excluded from consideration, the strain energy function depends on the state of deformation alone through the deformation gradient $F$. Let $\mathbf{X}$ be the coordinate of a point in the reference configuration and $\mathbf{x}$ its coordinate in the current configuration. The deformation gradient is given by

$$F = \frac{\partial \mathbf{x}}{\partial \mathbf{X}}.$$  \hspace{1cm} (1)

**Uniaxial Deformation.** In the case of uniaxial deformation, the deformation gradient is a diagonal tensor given by\textsuperscript{19}:

$$F = \begin{bmatrix} \lambda & 0 & 0 \\ 0 & \lambda^{-\frac{1}{2}} & 0 \\ 0 & 0 & \lambda^{-\frac{1}{2}} \end{bmatrix}.$$  \hspace{1cm} (2)

where $\lambda$ is the (applied) stretch in the axial direction. The right Cauchy strain tensor $\mathbf{C} = F^T F$ has the principal invariants $I_1 = \lambda^2 + \frac{2}{\lambda}$, $I_2 = 2\lambda + \frac{1}{\lambda^2}$, and $I_3 = 1$.

The nominal stress or the first Piola-Kirchhoff stress $T$ for uniaxial compression is given by\textsuperscript{19}:

$$T = \frac{2}{\lambda} \frac{\partial W}{\partial I_1} \left( \lambda^2 - \frac{1}{\lambda} \right) + \frac{2}{\lambda} \frac{\partial W}{\partial I_2} \left( \lambda - \frac{1}{\lambda^2} \right)$$  \hspace{1cm} (3)

where $W$ is the strain energy density function.

**Constitutive Material Models.** Based on prior experience and preliminary results from experiments, the exponential form of the strain energy function is chosen to represent the
mechanical behavior of bovine liver tissue at finite strains:

\[ W = B_1 \exp(B_2(I_1 - 3) - 1) \]  

(4)

where \( B_1 \) and \( B_2 \) are material parameters to be determined experimentally. The parameter \( B_1 \) has the dimensions of stress, while \( B_2 \) is a dimensionless parameter that controls the degree of material nonlinearity. From Eq. (3) and Eq. (4), the nominal stress for the exponential model for uniaxial compression is

\[ T = \frac{2}{\lambda - 1} \lambda B_1 B_2 \exp\left(B_2 \left(\lambda^2 + \frac{2}{\lambda} - 3\right)\right). \]  

(5)

The parameters \( B_1 \) and \( B_2 \) are found by curve-fitting Eq. (5) to experimental data from frictionless uniaxial compression tests. For a more detailed treatment of continuum mechanics, see\textsuperscript{19}.

### B. Correction Factors for the Exponential Material Model

The correction factor approach is based on the hypothesis that the error in the material parameters extracted from no-slip uniaxial compression experiments depends on the real material parameters as well as the diameter-to-height \((d/h)\) ratio of the samples. We determine this dependence for exponential hyperelastic materials through a detailed computational analysis of no-slip uniaxial compression experiments.

For a no-slip uniaxial experiment, we define the apparent material properties as those that satisfy

\[ T_{\text{app}} = \frac{F}{A} = 2 \left(\lambda - \frac{1}{\lambda^2}\right) B_{1,\text{app}} B_{2,\text{app}} \exp\left(B_{2,\text{app}} \left(\lambda^2 + \frac{2}{\lambda} - 3\right)\right) \]  

(6)

where \( F \) is the reaction force, \( A \) the sample’s initial cross-sectional area, and the subscript “app” emphasizes the fact that these are apparent properties determined from no-slip experiments. The quantity \( \lambda \) is understood as the ratio of the deformed height of the sample to its original height. We suppose that the ratios of the apparent properties to the true properties are functions of the true properties and the quantity \( d/h \):

\[ \frac{B_{1,\text{app}}}{B_1} = f(B_1, B_2, d/h) \]

\[ \frac{B_{2,\text{app}}}{B_2} = g(B_1, B_2, d/h). \]  

(7)
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The functions $f$ and $g$ are referred to as correction factors. If $f$ and $g$ are determined computationally, then Eq. (7) may be used to determine the material parameters directly from no-slip uniaxial experiments.

The functional forms of $f$ and $g$ are determined through the following steps:

1. Determine a range for the parameters $B_1$, $B_2$ and $d/h$ (usually through experiments).

2. For the selected parameter range, carry out FE simulations of no-slip experiments using the methods described in Section III. For each simulation, i.e., for fixed values of $B_1$, $B_2$ and $d/h$, compute the reaction force-stretch ($F - \lambda$) data and fit to Eq. (6) to extract $B_{1,\text{app}}$ and $B_{2,\text{app}}$.

3. Parametrize $f = B_{1,\text{app}}/B_1$ and $g = B_{2,\text{app}}/B_2$ in terms of $B_1$, $B_2$, and $d/h$.

With $f$ and $g$ in hand, the real material parameters $B_1$ and $B_2$ can be extracted directly from a no-slip uniaxial experiment using Eq. (6) by minimizing the quantity

$$
\sum_{i=1}^{N} \left[ \frac{F(\lambda_i)}{A} - 2 \left( \lambda_i - \frac{1}{\lambda_i^2} \right) f(B_1, B_2, d/h) \times B_{1,\text{app}} B_{2,\text{app}} g(B_1, B_2, d/h) \times B_2 \exp \left( g(B_1, B_2, d/h) \times B_2 \left( \lambda_i^2 + \frac{2}{\lambda_i} - 3 \right) \right) \right]^2
$$

where $N$ is the number of data-points in the experiment.

Strictly speaking, the correction factors are also functions of the maximum compression achieved in the uniaxial test. This depends on the application towards which the test is geared. We use a range of 30% compression in this work.

III. MATERIALS AND METHODS

A. Finite Element Modeling of the No-Slip Experiments

Finite element analyses of the no-slip uniaxial compression test were conducted using the commercially available software ABAQUS\textsuperscript{20}. As seen in Fig. 2, only a quarter of the cross-section of the platens and the tissue was included in the FE model to exploit symmetry and thereby reduce computational cost. Convergence tests were carried out to determine an
appropriate mesh size. The compression platen was modeled as an analytically rigid solid. A second-order axisymmetric hybrid element with reduced integration formulation was used to eliminate issues regarding volumetric locking due to the assumed incompressibility of liver tissue. The exponential material model was implemented with a user subroutine, UHYPER. The loading was applied incrementally as a displacement boundary condition on the top surface of the compression platen. Because the resulting stress distribution is non-uniform, the total reaction force $F$ was calculated at each step by numerically integrating the axial stress along the mid-plane (shown with a bold line in Fig. 2) with the appropriate scaling to account for the symmetries. The apparent nominal stress $T_{app}$ was then calculated as the total reaction force divided by the initial cross-sectional area and the apparent properties were found by fitting Eq. (6) to this data.

FIG. 2: Schematic of the model used in the finite element analysis of the no-slip compression experiment.

B. Experiments on Bovine Liver

We carried out two types of uniaxial compression tests on cylindrical samples of bovine liver tissue:

- No-slip experiments: The sample is perfectly bonded to the compression platens. This is the case from which one obtains the apparent material properties.
• Nearly frictionless experiments: The sample slides with almost no friction against the compression platens. This is the case from which one obtains the (almost) true material properties.

The results presented in this paper were obtained from a total of three bovine livers procured on three different days. Since sample preparation is a time consuming process, each liver was procured and tested on the same day to minimize the effects of ischemia on the material properties of the liver.

The experiments, schematically shown in Fig. 1, were conducted using a TestResources 100R Research Test System with a 2.2 lbf load cell. The livers were procured from local slaughterhouses at the time of slaughter and moved to the dissection laboratory at the University of Cincinnati in less than 30 minutes while stored on ice. There, each liver was sliced into thin layers parallel to the vertical line shown in Fig. 3(a). Cylindrical samples, with their axes parallel to the arrow in Fig. 3(a), were obtained with an 8 mm biopsy punch. The samples were immediately placed in cold phosphate buffered saline solution (PBS). The diameter of the experimental samples was constrained by the desire to obtain uniform liver tissue without blood vessels and other inclusions. The height of the samples was controlled by the thickness of the initial cut and the subsequent cuts with a surgical knife to create parallel surfaces. The samples were mostly extracted from the area to the right of the vertical line shown in Fig. 3(a) due to the more uniform nature of the tissue there.

![FIG. 3: Bovine liver: (a) Samples are obtained from the portion to the right of the vertical line. (b) A cylindrical sample at the end of a typical no-slip unconfined compression experiment.](image)

First, experiments with the nearly frictionless boundary conditions were conducted, during which liver tissue was allowed to contact the compression platens through a hydrated
layer, i.e., a nearly frictionless interface. The initial contact between the sample and the compression platen in these experiments was controlled manually to provide a 0.002 N pre-load. The compression experiment was initiated immediately after the pre-load was achieved to eliminate stress relaxation. A 0.01 s\(^{-1}\) strain rate was used for the nearly-frictionless experiments.

The experiments with the no-slip boundary condition were conducted with the use of Loctite Super Glue, which is a cyanoacrylate-based adhesive. The samples were bonded to cork pads, with adhesive backing, in order to increase the adhesive properties of the compression platens as well as to create a disposable surface; see Fig. 3(b). With the use of finite element modeling, the cork pads were confirmed to be analytically rigid relative to the soft liver tissue. Initially, one side of the sample was bonded to the cork that is attached to the lower compression platen. Next, the second cork pad was attached to the top compression platen. Once the glue was applied to the liver sample the top platen was lowered to contact the surface of the liver sample. The contact was controlled manually to result in a 0.002 N compression pre-load. As with the nearly-frictionless experiments, the uniaxial compression experiment was initiated quickly after the pre-load to avoid stress relaxation effects. The no-slip compression experiments were also conducted at a 0.01 s\(^{-1}\) strain rate.

IV. RESULTS AND DISCUSSION

A. Computational Determination of the Correction Factors \( f \) and \( g \)

To determine a parameter range for development of the proposed correction factors, we carried out nearly frictionless uniaxial compression experiments on bovine liver as described in Section III \((n = 13)\). The mean stress-stretch behavior from these experiments is shown in Fig. 4. A curve-fit of the exponential strain energy function to this data (shown in the figure) resulted in the parameters \( B_1 = 58.18 \) kPa and \( B_2 = 6.61 \).

Based on the nearly frictionless experiments, we selected the range \( B_2 = [5.0 : 0.3 : 7.7] \), i.e., ranging from 5.0 to 7.7 in increments of 0.3. The parameter \( B_1 \) was set to a constant value of 60.0 Pa because our FE analysis showed that the correction factors \( f \) and \( g \) are independent of \( B_1 \). The diameter was set to \( d = 0.01 \) mm and the height \( h \) ranged from
0.008 mm to 0.0144 mm in increments of 0.0008 mm. As a result, the diameter-to-height ratio ranged from 0.694 to 1.25. Each FE analysis simulated 30% compression of the sample.

![Graph showing stress-stretch results](image)

**FIG. 4:** Experimental stress-stretch results from the nearly frictionless unconfined compression experiments (n = 13). The exponential strain energy function provides an excellent fit to the data.

Figure 5 shows contour plots of the correction factors $f$ and $g$ as functions of $B_2$ and $d/h$. The error in the apparent properties increases as the parameter $B_2$ increases (i.e., as the material nonlinearity increases), and as the diameter-to-height ratio increases (i.e., as the area of no-slip increases). Note that $f < 1$ in the parameter range of interest, whereas $g > 1$. This means that the material parameter $B_1$ extracted from no slip experiments is always underestimated (by as much as 26%), while the parameter $B_2$ is always overestimated (by as much as 50%) in the parameter range of interest.

Figure 6 shows a contour plot of the product $f \ast g$. We consider this because the product $B_1 \ast B_2$ is proportional to the elastic modulus of the material\(^{21}\). Thus the quantity $f \ast g$ may be viewed as the correction factor for the material’s elastic modulus. The value of the function lies between 1.1 and 1.2, which means the elastic modulus determined from no-slip uniaxial compression experiments will always be 10%-20% above the true modulus (in the
FIG. 5: Contour plots of the correction factors $f$ (top) and $g$ (bottom) as functions of $d/h$ and $B_2$. 

Using the results of the parametric FE studies and analyzing the dependence of each correction factor on $B_2$ and $d/h$, we found that the following equations describe the correction factors accurately:

$$f(B_1, B_2, d/h) = \exp(0.024609 B_2 (d/h)^2)$$ (9)

and

$$g(B_1, B_2, d/h) = \exp(0.013844 B_2 + 0.2043389 (d/h)^2)$$ (10)

The term $d/h$ appears with a power of 2, suggesting that the area of contact is a critical parameter range considered here).
FIG. 6: Contour plot of the product of the correction factors $f$ and $g$, which may be considered the correction factor for the elastic modulus of the material.

B. Liver Material Parameters

In addition to the nearly frictionless experiments discussed in Fig. 4, we conducted 10 no-slip experiments on bovine liver tissue. Figure 7 shows three typical results from the no-slip experiments on bovine liver, each with a different $d/h$ ratio, and compares them to the mean response from the nearly frictionless experiments. Note that the stress at 0.70 stretch ratio for the no-slip case $d/h = 1.13$ is 60% more than that from the nearly frictionless response at the same stretch ratio. As expected, the deviation between the nearly frictionless and the no-slip stress-stretch responses increases as the $d/h$ ratio increases.

Next, in Fig. 8, we show the mean “corrected” response of bovine liver tissue, which is obtained as follows. We applied the proposed correction approach to the response from each of the 10 no-slip unconfined compression tests and determined the corrected material parameters $B_1$ and $B_2$ for each sample. Then, we calculated the stress-stretch response for each sample from the corrected parameters using Eq. (3) and computed the mean of these responses, which is shown in Fig. 8. The corrected data shows less spread than the nearly frictionless response in Fig. 4. From this mean corrected response, we found the
FIG. 7: Experimental stress-stretch results from the no-slip unconfined compression experiments. The solid line shows the considerably softer response from the nearly frictionless experiments.

corrected material properties of bovine liver at 0.01 s$^{-1}$ strain rate to be $B_1 = 101.17$ Pa and $B_2 = 4.81$.

The correction factor approach was validated against experimental data as follows. Using the corrected material parameters $B_1$ and $B_2$ specified above, we carried out FE simulations of no-slip compression experiments for two cases: $d/h = 1.13$ and $d/h = 1.28$. The resulting stress-stretch responses are compared with the responses from our no-slip experiments on bovine liver in Fig. 9. Very good agreement is observed between the experimental and computational responses for both values even though one of them ($d/h = 1.28$) lies outside the parameter range used to compute the correction factors. The figure also shows the corrected stress-stretch response of bovine liver and the experimentally determined nearly frictionless response. It is interesting to note that the corrected response is softer, suggesting the existence of some friction even in the nearly frictionless experiments.

Finally, we apply the proposed approach to the data of Chui et al.$^{16}$, wherein the authors determine the nonlinear material parameters of porcine liver from no-slip uniaxial experiments. These are apparent parameters because the no-slip boundary condition is not
FIG. 8: Mean stress-stretch curve obtained using the correction factor approach. The spread in the corrected response is much smaller than that in the nearly frictionless response.

accounted for in their analysis. We assume that our correction factors are applicable to their data. In Fig. 10, the open circles show the data from Chui et al. and fitting the exponential material model directly to this data (solid line) leads to the apparent properties $B_{1,\text{app}} = 59.52$ Pa and $B_{2,\text{app}} = 4.28$. The dashed line shows the corrected stress-stretch response, which is considerably softer; the corrected parameters are $B_1 = 66.98$ Pa, $B_2 = 2.96$. Using the corrected material parameters, we performed an FE analysis with no-slip boundary conditions, which results in the dotted line and shows very good agreement with the original no-slip data. We also applied the approach of Miller\textsuperscript{12} to correct the stretch data, which gives the response shown using triangles. We believe the lack of agreement between our correction and Miller’s correction is because the latter does not explicitly depend on the parameter $d/h$, which in our view plays a critical role in no-slip experiments.
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V. CONCLUDING REMARKS

In uniaxial compression experiments on soft tissue, friction between the platens and the sample is inevitable and very difficult to quantify. Bonding a sample to the platens eliminates the issues regarding unknown boundary conditions, but introduces a new mechanics problem with a non-uniform state of stress. Ignoring this fact can lead to significant overestimation of tissue stiffness. The proposed correction factor approach is able to account for the no-slip boundary condition and can be extended to other strain energy density functions of interest.

It is important to develop and use the correction factors in the context of practical experimentation. As the $d/h$ ratio increases, the surface area of the contact increases, which results in greater stiffening effects. Theoretically, one could decrease the $d/h$ ratio in order to minimize frictional effects. However, tissues such as liver are very soft and keeping a soft and tall sample upright is not always feasible. Also, the elimination of inclusions and

FIG. 9: Validation of the correction factor approach. Stress-stretch responses computed using the corrected material parameters and no-slip boundary conditions show good agreement with results from no-slip compression experiments on bovine liver tissue.
FIG. 10: Proposed correction approach applied to porcine liver data from Chui et al.\textsuperscript{16}. The corrected response is significantly softer than the no-slip data.

Other material imperfections becomes more difficult as the height of the sample increases. In light of these constraints, sample dimensions will inevitably result in frictional effects in unconfined compression experiments. It is therefore important to address the issue of friction in some manner. Toward this end, we propose adhering the sample to the compression platens and then appropriately correcting the response.

There are four major motivations for the characterization of soft tissue in general and liver tissue in particular: disease diagnosis, tissue engineering, injury prevention, and automated surgery. While each one of these applications has a unique operating strain rate, the compressive nature of loads is common to all of them. In this study, a strain rate of 0.01 s\textsuperscript{−1} is chosen so that we may reasonably assume the mechanical behavior of liver tissue to be hyperelastic. The capacity of the load cell used in this work was also a consideration in our choice of strain rate.

While this study focuses primarily on extracting the material parameters of tissue assum-
ing hyperelastic behavior, it is important to note that liver tissue is a visco-elastic material. Future work on this correction approach must therefore account for strain-rate dependence. In particular, for applications such as injury mechanics, one needs to consider much higher strain rates than $0.01 \text{s}^{-1}$. Moreover, it is necessary to take into account not just the visco-elastic nature of soft tissue but also possible damage to tissue.

Another limitation of this work is that the correction factors depend on the selected parameter range (material parameters, geometrical parameters, % compression). The larger this range, the wider the applicability of the correction factors. However, this is not a major limitation since the finite element simulations used to determine the correction factors need be performed only once for each point in the parameter space.

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FIG. 1: Schematic of the uniaxial compression experiment.

FIG. 2: Schematic of the model used in the finite element analysis of the no-slip compression experiment.

FIG. 3: Bovine liver: (a) Samples are obtained from the portion to the right of the vertical line. (b) A cylindrical sample at the end of a typical no-slip unconfined compression experiment.

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FIG. 5: Contour plots of the correction factors \( f \) (top) and \( g \) (bottom) as functions of \( d/h \) and \( B_2 \).

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