Frictional Force Modeling Ranging from Hyper to Slow Relative Velocity between a Needle and Liver Tissue*

Yo KOBAYASHI**, Takahiro SATO***, Takeharu HOSHI**
and Masakatsu G. FUJIE**
** Faculty of Science and Engineering, Waseda University,
59-309, 3-4-1 Ohkubo, Shinjuku-ku, Tokyo 169-8555, Japan
E-mail: you-k@fuji.waseda.jp
*** Graduate School of Science and Engineering, Waseda University,
59-309, 3-4-1 Ohkubo, Shinjuku-ku, Tokyo 169-8555, Japan

Abstract
Needle insertion treatments require accurate placement of a needle tip into the target cancer. However, it is difficult to insert the needle into a lesion because of tumor displacement during organ deformation. Therefore, path planning using a needle insertion simulation to analyze deformation of an organ is important for accurate needle insertion. A frictional model for needle insertion simulation is presented in this report. In particular, we focus on a model of frictional force based on the relative velocity between a needle and liver tissue ranging from hyper to slow velocity. In vitro experiments using porcine liver were performed at several relative velocities to measure the velocity dependence of the frictional force. Sixty trials of frictional force data were used to obtain average data at each relative velocity. The model of frictional force was then developed using the averages of the experimental results. This model was defined according to relative velocity, including hyper-slow velocity. Our modeling and experimental results show that the frictional force between the tissue and the needle increased during low relative velocity (under 1.5mm/s) and became constant (over 1.5mm/s).

Key words: Needle Insertion, Frictional Force, Liver, Velocity Dependence

1. Introduction
Percutaneous therapy has gained attention as a cancer treatment method. For example, percutaneous ethanol injection therapy (PEIT) and radio frequency ablation (RFA) are used to treat liver cancer. In this form of treatment, cancer cells existing inside the organ are necrotized by delivering a needle tip to the cancer cells to either inject ethanol (PEIT) or to ablate the cells (RFA). Percutaneous therapy has become a major trend in liver cancer treatment and has the advantages of being minimally invasive while providing acceptable results. In recent years, research and development have been performed with surgical robots and navigation systems for minimally invasive and precise surgery. Research into robotic systems to assist needle insertion has also been conducted to improve the accuracy of needle placement and expand the approach path. In percutaneous therapy, it is necessary to accurately place the needle tip at the target cancer. Since the target organ, such as the liver, consists of soft tissue, organ morphology can become deformed, which also changes the
position of the target lesion. Therefore, it is necessary to devise a plan for an insertion path that takes organ deformation into account. A numerical simulation in a virtual surgery environment reproduced with physical models of organs may be used to create such a plan.

A number of research groups have conducted studies on the development of a deformable organ model. Alterovitz et al. investigated the simulation of steerable needle insertion for prostate brachytherapy (1). DiMaio et al. developed a linear system for analyzing the extent of phantom deformation of planar tissue during needle insertion by using a linear elastic material model (2, 3). Dehghan et al. presented a planning system to determine the optimized insertion angle and position using nonlinear organ models (4).

A number of research groups have investigated force measurements during needle insertion as follows. Kataoka et al. investigated the relationship between needle deflection and force (5, 6). Abolhassani et al. evaluated the needle insertion force into turkey muscle tissue and presented an online updated algorithm of the needle trajectory (7). Studies by Okamura et al. developed empirical models for the needle insertion force, comprising tissue stiffness force, frictional force and puncture force (8, 9, 10). Heverly et al. showed the velocity dependency of the puncture force of a biomaterial (11). Podder et al. described the needle insertion force during prostate brachytherapy (12). Studies by Barbe et al. investigated an online model estimation of the needle insertion force (13, 14). We also previously developed and reported a biomechanical liver model, for which we provided specific descriptions of the material properties of the liver and finite element-based modeling, and presented a validation of the proposed model (15). Additionally, we reported a probability-based puncture condition (16), a planning method based on deformation simulation of the liver and a model of the puncture conditions (17).

Needle insertion forces are defined such that the stiffness force occurs before puncture of the capsule, and the frictional forces occur after this main puncture. This delineation suggests that detailed modeling of each component is necessary for path planning to achieve accurate needle insertion. An analysis of needle advancement into the liver should include the frictional force in the simulation, based on numerous reports that have revealed a correlation between frictional force and organ deformation. In the current study, we focused on the detailed modeling of frictional force between organ tissue and a needle (Fig. 1). A number of studies have discussed the frictional properties between needles and organ tissues during needle insertion, as well as the use of these properties for needle insertion. However, the frictional properties included in these models are a coarse approximation of the actual friction conditions. In particular, a problem is that many frictional models are defined considering only the velocity of the needle, whereas the magnitude of the frictional force is generally decided according to the relative velocity. Therefore, a model of frictional force should be defined based on the relative velocity between the needle and organ tissue. Moreover, frictional forces are distributed among each contact point of the needle, and therefore, the velocity of organ deformation varies depending on these contact points. The relative velocity near the needle tip is assumed to be low, while the relative velocity far from the needle tip is high. Specifically, the relative velocity at the needle tip is zero, except at the time of puncture. The frictional force near the needle tip is assumed to be very slow. Therefore, evaluation of the distributed frictional force at each contact point must correspond to a relative velocity from a hyper-slow range.

In this study, we focused on the measurement and modeling of frictional force based on relative velocity, including hyper-slow velocity. Okamura et al. have already measured the frictional force according to the relative velocity (10). However, the frictional force near zero relative velocity was not accurately measured in their study, because of the manual segmentation of liver tissue and a needle from CT images. Therefore, we aimed to determine the biomechanical properties of the frictional force near zero relative velocity. We also describe the methods used to measure frictional force and velocity dependence, discuss
the modeling of frictional force considering velocity dependence and time varying properties, and validate the proposed frictional model through an in vitro experiment.

![Diagram of needle insertion and force conditions](image1)

**Fig. 1** A schematic of needle insertion and force conditions

## 2. Method

### 2.1 Liver condition

Porcine livers were cut into rectangles (thickness: 20 mm) and sandwiched by plastic plates (Fig. 2). The needle velocity was considered to be the same as the relative velocity, because the plates constricted liver deformation. This setup demonstrated the rigorous conditioning of the relative velocity between the needle and liver tissue.

### 2.2 Experimental equipment

A bevel-tip 17-gauge biopsy needle was used for the experiment. As shown in Fig. 2, the configuration of the experimental equipment was performed using a needle with a linear stage and a force sensor (LVS-200A, KYOWA Ltd.). The linear stage enabled a degree of freedom in the direction of needle insertion, and the force sensor enabled measurement of the force applied to the needle.

### 2.3 Experimental conditions

The experiments were conducted as follows (Fig. 2):

1) **Initial insertion**: The needle was inserted into the experimental sample of the sliced liver at a constant velocity of 1 mm/s. The needle stopped after penetration of the sample.

2) **Measurement of frictional force**: After the initial insertion, the needle was translated at a constant velocity, while the experimental conditions of needle insertion velocity were set to 30 patterns ranging from 0.01 mm/s to 10 mm/s. The needle was translated 40 mm during the experiments. The experiments were conducted with 24 porcine livers, and experiments were performed for each needle insertion velocity. Sixty trials were conducted for each insertion velocity. The force on the needle was measured during each experiment.

![Experimental setup and conditions to model frictional force](image2)

**Fig. 2** Experimental setup and conditions to model frictional force
3. Results and modeling

This section describes the experimental results for frictional force and its modeling based on the experimental results.

3.1 Results and modeling of frictional force

Figure 3a shows representative experimental results of frictional force when the needle insertion velocity was 10 mm/s. Each experimental result showed an increase in frictional force with time. The results of the three experiments were confirmed to have variance, even though the results were obtained under the same experimental conditions. To reduce this variance, a total of 60 sets of data of frictional force were used to obtain average data. Figure 3b shows a representative example of the average frictional force when the needle insertion velocity was 10 mm/s. We then modeled this experimental result of frictional force on the needle using the following equation:

\[ F_{friction} = F_o + at \]  

(1)

where \( F_{friction} \) is the frictional force on the needle, \( F_o \) is the initial frictional force, \( a \) is a coefficient indicating the rate of increase in force, and \( t \) is time.

Fig. 3 Representative result of frictional force (a) and the average frictional force (b). Error bars indicate standard deviation.
3.2 Modeling of velocity dependence

We discuss below the frictional force using relative velocity, instead of needle insertion velocity, since the relative velocity between the liver and needle was the same as the needle insertion velocity in this experiment.

I) Initial frictional force $F_o$: The initial frictional force $F_o$ at each relative velocity is summarized by the same process described in §3.1. Figure 4a shows the results of velocity dependence of the initial frictional force $F_o$. A semi-logarithmic graph of the same data is shown in Fig. 4b. A change in tendency was observed at a level near the relative velocity at 1.5 mm/s (Fig. 4a). Additionally, the initial frictional force $F_o$ was considered to exhibit low-velocity characteristics when the relative velocity was lower than 1.5 mm/s and high-velocity characteristics when the relative velocity exceeded 1.5 mm/s. Therefore, the velocity dependence of initial frictional force $F_o$ was investigated using each set of results. The results displayed in Fig. 4b indicate that the initial frictional force $F_o$ increased proportionally during the low-velocity range. We then modeled low-velocity characteristics using Eq. (2). We found that the high-velocity frictional force was approximately constant (Fig. 4a). We then modeled the high-velocity characteristics using the following equation (3).

$$F_o = A \ln(v) + B \quad (2)$$
$$F_o = F_s \quad (3)$$

where $A$ and $B$ are parameters, $\ln()$ is the logarithm function, $v$ is the relative velocity between the liver tissue and the needle and $F_s$ is high-velocity frictional force.

II) Rate of increase in force $a$: The rate of increase in force $a$ at each relative velocity was averaged by the process described in §3.1. Our results indicated that the rate of increase in force $a$ increased proportionally as relative velocity increased (Fig. 5). We then modeled the rate of force increase $a$ using the following equation:

$$a = K_v \quad (4)$$

where $K$ is a parameter indicating the slope of the line shown in Fig. 5.

The frictional model depending on the relative velocity was modeled based on the above discussions using the following equation:

$$F_{friction} = \begin{cases} 
A \ln(v) + B + K_v t & (0.01 < v < 1.5) \\
F_s + K_v t & (v \geq 1.5)
\end{cases} \quad (5)$$

Table 1 shows each value of the parameter.
Fig. 4 Velocity dependence of the initial frictional force as a linear graph (a) and as a semi-logarithmic graph (b).

Fig. 5 Velocity dependence of the rate of increase in force \( a \).

Table 1 Parameters of Eq. (5)

<table>
<thead>
<tr>
<th>A [Ns/mm]</th>
<th>B [N]</th>
<th>K [N/mm]</th>
<th>( F_s ) [N]</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.046</td>
<td>0.51</td>
<td>0.0021</td>
<td>0.55</td>
</tr>
</tbody>
</table>
3.3 Evaluation of parameter variance

As shown in Fig. 4, there were significant variations in the initial frictional force $F_o$. Variations emerged even when the needle insertion experiments were conducted under virtually identical conditions. Therefore, the conditions under which the frictional force may be determined by the micro status of the needle and the tissue were considered. It is important to measure variations in the experimental data from a probabilistic point of view. A histogram of the data shown in Fig. 6 was developed to visualize the status of the distribution of the initial frictional force $F_o$ when the relative velocity was 5 mm/s. In Fig. 6, the Y axis on the left-hand side represents the number of samples in the interval, while the Y axis on the right-hand side represents the discrete probability distribution, where the latter value was calculated using the following equation:

$$ p_i(i) = \frac{n_i}{N\lambda_i} $$  \hspace{1cm} (6)

where $n_i$, $\lambda_i$ and $N$ represent the number of data for the $i$ interval, the space of the $i$ interval and the number of total data, respectively.

The probability distribution of the initial frictional force $F_o$ was modeled based on the histogram shape, which has the following features:
- The initial frictional force was defined only in the non-negative region.
- The histogram in Fig. 6 had a distorted distribution.

Based on the above features, the gamma distribution shown in the following equation was employed as a model representing the puncture probability distribution $p$:

$$ p(F_o) = \frac{F_o^{\alpha-1}}{\beta^n \Gamma(n) \exp(-\frac{F_o}{\beta})} $$ \hspace{1cm} (7)

where $F_o$ is the initial frictional force; $\alpha$ and $\beta$ are parameters used to determine the shape of the gamma distribution, and $\Gamma$ is the gamma function. In the gamma distribution, the median $\mu$ and variance $s$ in terms of $\alpha$ and $\beta$ are as follows:

$$ \mu = \alpha \beta, $$  \hspace{1cm} (8)
$$ s^2 = \alpha \beta^2. $$  \hspace{1cm} (9)

Figure 6 shows the probability distribution modeled by the gamma distribution. The values of parameters $\alpha$ and $\beta$ in Table 2 were determined using Eqs. (8) and (9) from the values of the median $\mu$ and variance $s$. Figure 6 also shows a histogram of the status of the distribution, and the probability distribution model with a gamma distribution shows a similar tendency. This result confirmed the capability of the gamma distribution to model the probability distribution of parameters, representing the variation of the experimental data.
4. Evaluation experiment

This section describes the experiment used to evaluate the model of frictional force shown in §3. The relative velocity between the needle and the liver tissue, and the frictional force were measured. We discuss the model of frictional force based on the experimental results below.

4.1 Experimental method

The liver was cut into a rectangular section (thickness: 20 mm), and both sides of the liver were secured as fixed ends. Both sides of the liver were attached to sandpaper using glue, and the sandpaper was attached to the wall by double-sided tape. No rearward securing was performed (Fig. 7a).

For the measurement of the relative velocity between the liver tissue and the needle, a marker was attached near the contact point of the needle and the liver tissue (Fig. 7a). The position of the marker was measured, and its position was used to determine the position of the contact point through image recognition using a camera image, which was set up on the upper side of an experimental table. The position of the marker was used to calculate the relative velocity between the needle and the liver tissue. The experiments were conducted as follows:

1) Initial insertion: The needle was inserted into the experimental sample of the sliced liver at a constant velocity of 1 mm/s. The needle stopped after penetration of the sample.

2) Measurement of frictional force: After the initial insertion, the needle was translated at a constant velocity of 5 mm/s. A bevel-tip 17-gauge biopsy needle was used for the experiment. This is the same needle used in the experiment shown in §2. The axial force on the needle was measured during needle insertion. The force on the needle and the marker position were measured during the experiment.

Table 2 Parameters of Eqs. (7) ~ (9)

<table>
<thead>
<tr>
<th>$u$ [N]</th>
<th>$s$</th>
<th>$\alpha$</th>
<th>$\beta$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.56</td>
<td>0.043</td>
<td>165</td>
<td>0.0034</td>
</tr>
</tbody>
</table>

Fig. 6 Histogram of the initial frictional force. The Y axis on the left-hand side represents the number of samples in the interval, whereas the Y axis on the right-hand side represents the discrete probability distribution. The red line represents the model of the gamma distribution.
4.2 Results

Figure 7b shows the experimental data of the liver position, relative velocity between the liver tissue and the needle, and frictional force. The experimental results obtained during the early stage of the experiment ($t < 5\ \text{s}$) showed that the frictional force increased nonlinearly, whereas the liver position increased linearly at the same velocity as the needle, indicating that the relative velocity was approximately zero. In this case, the nonlinear increase in frictional force was considered to correspond to the increase in relative velocity, while this was not confirmed for the low resolution data of the relative velocity because of low resolution and a low update time of the camera image. It is suggested by the data shown in Fig. 4a, which show a rapid increase in frictional force during low relative velocity, ranging up to $1.5\ \text{mm/s}$. The experimental results during the later stage of the experiment ($t > 5\ \text{s}$) showed that the rate of increase in the frictional force was reduced compared with the data during the early stage ($t < 5\ \text{s}$), while the relative velocity was greatly increased. In this case, the increase in force ceased, because the relative velocity exceeded the slow velocity range ($v < 1.5\ \text{mm/s}$).

![Fig. 7 Setup (a) and results (b) of the evaluation experiment.](image)
5. Discussion

5.1 Modeling

In the current study, we compared our model to well-known existing friction models. For example, the Coulomb model includes constant dynamic friction and static friction at zero relative velocity, whereas the Karnopp friction model includes dynamic friction and static friction within a “dead zone” near zero velocity (19). Models, such as the Stribeck effect, show that the value of the friction force decreases as velocity increases at low velocities (20). The Dahl model includes pre-sliding displacement (21). The frictional model in the existing models consists of static friction at zero relative velocity and dynamic friction at non-zero relative velocity. The frictional force in our model can be separated into “static friction” at low relative velocity (<1.5 mm/s) and constant dynamic friction at high relative velocity (>1.5 mm/s). This distinction between our experimental results and other models suggests that the mode shift in frictional force cannot be separated at zero, but it can be separated at a certain relative velocity instead. Collectively, these results demonstrate that frictional force is increased at a low relative velocity, which may be considered to be almost zero in conventional situations as relative velocity is increased.

5.2 Evaluation experiment

As described in the results, the tendency of frictional force changed when the relative velocity between the needle and the liver tissue exceeded approximately 1.5 mm/s. Additionally, the experimental results showed a rapid increase in frictional force during low relative velocity (v < 1.5 mm/s) and a low rate of increase in frictional force during high relative velocity (v > 1.5 mm/s). This condition is the same as that in the model of frictional force proposed in Eq. (5), supporting this frictional model.

5.3 Limitations

It is important to refine the design of the system, considering its use in clinical practice. The limitations of the present work are described below.

1) Evaluation of accuracy: In spite of a real-time visible image from an ultrasound probe, accurate placement of the needle is still difficult, especially with small, early tumor lesions that require placement accuracy of only a few millimeters. Because the liver is soft, it can easily be deformed when a needle pierces it and advances toward an inner tumor. The deformation causes displacement of the tumor lesion. We also previously developed and reported a biomechanical liver model, for which we provided specific descriptions of the material properties of the liver and finite element-based modeling, and presented a validation of the proposed model (15). Additionally, we reported a probability-based puncture condition (16), a planning method based on deformation simulation of the liver and a model of the puncture conditions (17). In this planning method, we simulated liver deformation before the needle punctured the tissue surface and frictional force was not considered. The needle insertion experiment demonstrated that the proposed planning method actualized a needle placement accuracy of approximately 1.5 mm. To improve the 1.5 mm needle placement accuracy from our previous validation experiment (17), an analysis of needle advancement into the liver should include the frictional force in the simulation, based on numerous reports that have shown a correlation between frictional force and organ deformation. DiMaio et al. developed a finite element analysis (FEA) during needle insertion, which included tissue deformation analysis with needle-tissue frictional interaction (2, 3). Our frictional model can be used for a more accurate path planning method integrated with such a needle-tissue frictional interaction model for future
Intraoperative parameter identification: We observed a large variance in frictional force (Fig. 6). It was assumed that these differences resulted from the fact that the parameters of preoperative planning are different from those of the actual liver. We believe that it is necessary to take into account modeling errors caused by parameters to estimate the insertion accuracy more precisely. In future studies, we plan to employ a method that uses intraoperative information to identify parameters (22, 23). Use of the parameter identification method may improve the accuracy of estimation.

III) Difference between experimental conditions and actual operation procedure: One of the limitations of the present study is the difference between experimental conditions and the actual operation procedure. The needle penetrated the liver tissue in our experiment, but the needle does not penetrate the liver and the needle tip stays inside the liver in the operation procedure. This difference may affect the phenomenon and modeling result. This issue will be investigated in further studies.

IV) Effect of contact length: The above experimental results were obtained using 20-mm-thick sections of porcine liver. The frictional force was affected by the contact length between the liver tissue and the needle. Therefore, the modeling of frictional force considering the contact length should be investigated in future work.

V) Evaluation experiment: the lack of quantitative investigation to evaluate our frictional model is one of the limitations of the present study because our experimental results only showed the qualitative correspondence of the frictional phenomenon between the model and the experiment. As described above, FEA, which includes tissue deformation analysis with needle-tissue frictional interaction, should be carried out in future studies for more robust evaluation.

6. Conclusions and future work

In the present study, we described a frictional model for needle insertion simulation. We focused on a frictional model based on the relative velocity between the needle and liver tissue, including a hyper-slow velocity. In vitro experiments using porcine liver were performed at several relative velocities to measure the velocity dependence of frictional force. Sixty sets of frictional force data were used to obtain average data for each needle insertion velocity. A frictional force model was then developed from the experimental results. The proposed model was defined using the relative velocity and a range from hyper-to slow velocity. Finally, an evaluation experiment was performed, showing that the frictional force changed corresponding to the relative velocity between the needle and liver tissue. Therefore, the experimental results support the proposed frictional model.

In future studies, the frictional force model will be used in needle insertion simulations and a planning method will be proposed to achieve a more accurate needle path. Moreover, a model parameter identification method using intra-operative information will be investigated, with reference to ambiguities of the model parameters from individual differences of patients.

Acknowledgments

This work was supported in part by the "Establishment of Consolidated Research Institute for Advanced Science and Medical Care", the Encouraging Development Strategic Research Centers Program, and the Special Coordination Funds for Promoting Science and Technology from the Ministry of Education, Culture, Sports, Science and Technology; the Global Centers of Excellence Program “Global Robot Academia,” of Waseda University, Tokyo, Japan; and in part by a Grant-in-Aid for Young Scientists (B) (21700513) and a
Grant-in-Aid for Scientific Research (B) (22360108) from the Ministry of Education, Culture, Sports, Science and Technology of Japan.

References


