Effects of Needle Inner Surface Topography on Friction and Biopsy Length

Weisi Li^{1,2}, Ping Zhou¹, Wei-Chen Lin², Valens Nteziyaremye³, Hitomi Yamaguchi³,
Dongming Guo¹, Albert Shih²

¹ School of Mechanical Engineering, Dalian University of Technology, Dalian, Liaoning 110042, China

² Department of Mechanical Engineering, University of Michigan, Ann Arbor, Michigan 48109, USA

³ Department of Mechanical & Aerospace Engineering, University of Florida, Gainesville, Florida 32611, USA

Abstract

The effects of inner surface topography on friction force and the coefficient of friction between an inner surface of needle and tissue phantom during needle insertion and on biopsy length are examined using polished, and unpolished needles. In this paper, the inner surfaces of the needles are altered by magnetic abrasive finishing (MAF), and the needle inner surfaces were smoothed from 0.43 μm to 0.02 μm *Sa* (referred to as the *smoothly polished needle*) or 0.26 μm *Sa* (referred to as the *roughly polished needle*). With polyvinyl chloride (PVC) as a tissue-mimicking phantom material, the smoothly polished needle results in the smallest inner friction force (0.0024 N/mm), smallest coefficient of friction (0.35), and the longest biopsy length (34.3 mm). The roughly polished needle produces the largest inner friction force (0.0041 N/mm), the largest coefficient of friction (0.56), and the shortest biopsy length (30.5 mm). Although the inner surface roughness of the roughly polished needle is lower than that of the unpolished needle, the inner friction force and coefficient of friction are larger than those of the unpolished needle. This may be caused by the ridges and grooves in the circumferential direction of the roughly polished needle made with MAF. Based on the surface topography data, the ratchet mechanism is utilized to calculate the coefficient of friction. The results suggest a higher coefficient of friction in the roughly polished needle. The biopsy tests also show the shortest biopsy sample is extracted using the roughly polished needles, which exhibit the greatest inner friction force.

Keywords: needle, magnetic abrasive finishing, friction force, biopsy length

1. Introduction

Advances in medical imaging, particularly magnetic resonance imaging (MRI), can detect cancer in early stages. This is often followed by a minimally invasive needle procedure to acquire tissue samples for histological diagnosis. For example, prostate cancer patients under the active surveillance protocol require the prostate needle biopsy every 6 to 12 months to monitor the progress of the cancer [1–4]. The needle biopsy tissue length greatly influences the prostate cancer detection rate [5–8]. Iczkowski *et al.* [5] observed an increasing trend in the probability of cancer detection with biopsy core length ranging from 1 mm to 28 mm. Öbek *et al.* [6] found that the prostate cancer detection rate with biopsy core length longer than 11.9 mm was 39%, higher than 23% for cases shorter than 11.9 mm. Yilmaz *et al.* [7] detected 97.4% prostate cancer in biopsy cores longer than 6 mm, which was higher than 71.8% in shorter biopsy samples. Fink *et al.* [8] found that increasing the needle biopsy tissue length from 19 mm to 29 mm raised the prostate cancer detection rate by 18%. To reduce the false negative diagnosis and pain caused by repeated prostate biopsies, the length of tissue acquired in each needle biopsy needs to be increased.

Needle tubes are fabricated by the drawing process. While drawing the small diameter, thin wall needle tubes, the process is limited to use only the outer die without the inner mandrel [9]. Therefore, the inner surface of the needle tube is rough. The commercial needle in a study [10] showed the inner surface roughness was $0.43 \mu\text{m Sa}$, almost twice higher than the $0.23 \mu\text{m Sa}$ of the outer surface of the same needle. Such rough inner surface is expected to affect the tissue length in needle biopsy. Magnetic abrasive finishing (MAF) has been demonstrated to be able to reduce the surface finish and friction force on needle inner surface [10,11] and potentially increase the tissue length in biopsy.

In hollow needle insertion to the soft tissue or tissue-mimicking phantom, as shown in Fig. 1(a), the measured insertion force is the sum of needle tip cutting force (F_c) and the inner (F_i) and outer (F_o) friction force. The cutting force is determined by the needle tip geometry. Moore *et al.* [12,13] investigated the effect of the needle tip geometry on the cutting force and the biopsy length and concluded that smaller bevel angle led to lower cutting force and longer biopsy sample. The friction force during needle insertion has been measured in several studies [14–20]. The outside friction force F_o can be measured when the needle penetrates over the phantom tissue in Fig. 1(c). A method to measure inner friction force F_i between needle inner surface and phantom core during biopsy has been developed [10]. The F_i of needle inner surface with different levels of MAF and relationships between F_i and biopsy length have not been quantitatively investigated.

Figure 1. Needle inner friction force measurement with different length PVC phantoms.

The goal of this study is to investigate the effects of three needle inner surfaces, 1) smoothly polished with long polishing time, 2) roughly polished with short polishing time, and 3) as-received surface, on F_i and tissue length in needle biopsy using the soft tissue mimicking polyvinyl chloride (PVC) phantom. Methods for needle inner surface friction force measurement, prediction of normal stress inside the needle, discussing the effect of surface topography on friction based on the ratchet mechanism, and acquiring biopsy sample from the PVC phantom are first described. Experiments on needle surface polishing and needle insertion to measure inner friction force and biopsy length are then presented. Results of the inner friction force, coefficient of friction, and biopsy length for three types of needles are discussed. The effect of the surface topography of three surface types on friction were analyzed based on the ratchet mechanism to explain the experimental results.

2. Methods

2.1. The needle inner surface friction force measurement

The inner friction force measurement method used in [10] was adopted in this study. As shown in Fig. 1(b), at the moment when the needle penetrates the tissue, the insertion force F_1 is equal to $F_i + F_o + F_c$. After the needle has penetrated the same tissue, as shown in Fig. 1(c), the insertion force is $F_2 = F_o$. The difference of the insertion force after penetration is $F_1 - F_2$, which is equal to $F_c + F_i$. By measuring three tissue samples of different lengths (marked as l_1 , l_2 , and l_3 in Fig. 1(d)) as an example, the $F_1 - F_2 (= F_c + F_i)$ increased with a longer tissue length. Assuming the cutting force F_c remains the same during needle insertion, a line that fits these three data points will intersect the Y-axis ($l = 0$) at $(0, F_{c0})$. This approach linearly extrapolates the experimentally measured $F_c + F_i$ to estimate the value of F_{c0} when the tissue length is equal to zero. The F_i per unit length, denoted as f_{fi} in Fig. 1(d), is the slope of the best-fit line. Mathematically, the relationship between $F_c + F_i$ and l is

$$F_c + F_i = F_{c0} + f_{fi}l . \quad (1)$$

2.2. Normal stress of phantom tissue inside needle

The needle inner coefficient of friction (μ), which is the ratio of the friction vs. normal force. Based on the assumption that the distribution of the inner friction force and normal force is uniform for the whole inner surface, the μ is calculated with the inner friction force per unit length (f_{fi}) and inner normal force per unit length (f_{Ni}):

$$\mu = \frac{f_{fi}}{f_{Ni}} = \frac{\sigma_{fi}}{\sigma_{Ni}} , \quad (2)$$

where the σ_{fi} is the inner shear stress, which can be calculated by dividing the experimentally measured f_{fi} with the perimeter of the needle inner circle, and the σ_{Ni} is the inner normal stress on needle inner surface. The σ_{Ni} can be estimated by calculating the stress required to uniformly compress the tissue core in the radial direction from the original core diameter D_{co} to the compressed core diameter D_{cc} under the plane stress condition. Both D_{co} and D_{cc} are marked in Fig 2. Because the phantom core is very soft, it is very easy to deform. The deformation of the phantom core during measurement affects the core diameter significantly. The hole left in the PVC sample should have the same diameter (D_{co}) as the core which is removed from the sample. Furthermore, since the PVC phantom is transparent, the diameter of the hole made inside the

PVC sample can be measured nondestructively from outside. Because the diameter of the cylinder is much larger than the hole diameter, the cylinder surface is close to plane in the measuring location. The dimension distortion caused by the light refraction in the cylinder is very small and could be ignored in this study. Therefore, the diameter of the hole left in the PVC sample D_h was measured as the D_{co} in this study. A caliper was used to measure the distance between the extents of the hole as D_h . After the core enters the needle, the core has the same diameter as the inner needle diameter: D_{cc} is equal to the needle inner diameter D_i . D_i was measured optically by taking a magnified picture of the needle inner surface cross-section.

Figure 2. A schematic of the diameter reduction of tissue phantom core

In the cylindrical coordinate system, the general solution of the radial stress σ_r in the radial direction with distance r from the center is [21]:

$$\sigma_r = C_1 + C_2 r^{-2}, \quad (3)$$

where C_1 and C_2 are two constants to be determined by boundary conditions. The radial displacement, Δ , could be expressed as [21]:

$$\Delta = \frac{1}{E} [(1-\nu)C_1 r - (1+\nu)C_2 r^{-1}], \quad (4)$$

where E is the elastic modulus and ν is the Poisson's ratio. At $r = D_h/2$, Δ is $(D_h - D_i)/2$. At the center ($r = 0$), Δ is 0. Base on these two boundary conditions, $C_1 = \frac{E(D_h - D_i)}{(1 - \nu)D_h}$ and $C_2 = 0$. The σ_r could be expressed as [21]:

$$\sigma_r = \frac{E(D_h - D_i)}{(1 - \nu)D_h} . \quad (5)$$

The σ_r was uniform in the whole domain and $\sigma_{Ni} = \sigma_r$.

The ν of PVC was assumed to be 0.45 in this study [22]. The elastic modulus E of PVC could be approximately obtained with linear fitting of the measured stress-strain curve from compression test in the small strain region in previous study [23]. The E of the PVC in this study was 11.9 kPa.

2.3. Ratchet mechanism of surface topography on friction force

The ratchet mechanism [24] has been utilized to analyze the friction between two rough surfaces with asperities of one surface much smaller in lateral size than those of the mating surface. If one surface is more compliant than the other contact surface, the ratchet mechanism has a significant effect [25]. In this study, the PVC is much more compliant than the stainless steel needle. The global coefficient of friction between PVC and the needle inner surface is estimated based on the ratchet mechanism and surface topography data.

In ratchet mechanism, assuming the adhesive component of the coefficient of friction is constant (μ_0). If the local normal force is N , the local tangential friction force T is equal to $\mu_0 N$. Experimentally, the measured coefficient of friction in the smoothly polished surface can represent μ_0 . The coefficient of friction along the moving direction (μ_m) varies with the topography of the surface.

On a rough surface as shown in Fig. 3, the force along (F) and vertical (W) to the moving direction could be expressed with T , N and the angle of the slope with the horizontal plane θ .

$$W = N \cos \theta - T \sin \theta . \quad (6)$$

$$F = N \sin \theta + T \cos \theta . \quad (7)$$

The measured local μ_m is equal to F/W . Based on Eqs. (6) and (7), the μ_m was

$$\mu_m = \frac{F}{W} = \frac{\mu_0 + \tan \theta}{1 - \mu_0 \tan \theta} . \quad (8)$$

The μ_m is dependent on θ , which can be calculated using the surface topography data. The measured coefficient of friction (μ_a) of the overall surface is the average of μ_m across the whole surface and influenced by the surface topography.

Figure 3. Schematic of the effect of the rough surface topography on the apparent coefficient of friction μ_a .

The surface topography data consists of the height of $P \times Q$ points, where P and Q are the numbers of points along and across the sliding direction, respectively. The line segment between two adjacent points (z_i, y_i) and (z_{i+1}, y_{i+1}) along a line in the sliding direction has an angle θ_i , as shown in Fig. 3. Region I is the upslope segment and has a positive θ_i (Eq.(9)). Region II is the downslope segment with a negative θ_j which could be calculated similar to Eq. (9).

$$\theta_i = \arctan \frac{y_{i+1} - y_i}{z_{i+1} - z_i} . \quad (9)$$

Assuming the μ_0 is constant for the whole surface, based on Eq. (8), the μ_m of each segment could be calculated with the local θ_i and the given μ_0 . During the sliding, the upslope region impeded the PVC material to move forward while the downslope region did not. In addition, since the PVC phantom is soft and had large deformation in the advancing direction, most parts of the PVC phantom contacted with the upslope region. Therefore, the contact force mainly distributed on the upslope ($\theta_i > 0$) region, as Region I shown in Fig. 3. It is assumed that the coefficient of friction for the whole surface μ_a is the result of the sum of coefficient of friction (μ_{up}) in the upslope segments (Region I) divided by the number of upslope segments, N_{up} .

$$\mu_a = \frac{\sum \mu_{up}}{N_{up}} . \quad (10)$$

2.4. Separating the end of biopsy sample from the PVC phantom

The method to cut and separate the long, thin, and soft biopsy PVC phantom inside the needle is shown in Fig. 4. The biopsy needle is first inserted into the phantom with a given depth (30 mm in this study). The needle rotates around its axis (Fig. 4(a)) to cut the material in the circumferential direction at the needle tip. The phantom is tilted by about 5° and the needle tip cuts and separates the base of core inside the needle (Fig. 4(b)). Finally, the needle advances by about 10 mm further to cut off the biopsy core (Fig. 4(c)). When the needle retracted from the phantom, the biopsy core is inside the needle and can be pushed out with the assist of a stainless steel wire (thinner than the needle inside diameter). The length of the biopsy core is measured using a caliper.

Figure 4. Schematic of the biopsy core acquisition method in PVC phantom: (a) the needle is inserted into the phantom and rotates, (b) the phantom tilts and needle tip cuts the end of the PVC core inside the needle, and (c) the needle advances further to cut off the core.

3. Experimental Setup

3.1. Needle MAF polishing

The needles used in this study were 18 gauge (1.27 mm OD and 1.08 mm ID) AISI 316 stainless steel needles. The surface roughness of the as-received inner surface was 0.43 μm S_a (4.14 μm S_z). The MAF experimental setup, as shown in Fig. 5, was utilized to polish the inner and outer surfaces of the needles. Two magnetic poles with 90° configuration were driven by the linear slide to feed in the axial direction of the needle with 0.59 mm/s. The needle tube was

chucked to a motor and placed between the two magnet tips. The magnetic field produced by the magnetic tip pushed the magnetic abrasives onto the tube inner and outer surface and provided the force to make the abrasives scratch on the surface when the tube was rotating at 10,000 rpm. The magnetic abrasives used in this paper were the mixture of 80 wt% iron particles (150 to 300 μm diameter) and 20 wt% aluminum oxide abrasive (80 μm mean diameter). With longer polishing time (10 polishing passes), the needle inner surface could be smoothly polished. With only two polishing passes, the needle inner surface was roughly polished. The outer surfaces of three types of needles were all polished by MAF to be with the same surface roughness. The trial polished needle was sectioned to expose the inner surface for roughness measurement using a white light interferometer (NewViewTM 7200, Zygo, Middlefield, CT). After the needle tube was polished, the needle tip was ground with a 3-axis surface grinding machine and a 2-axis needle fixture [26]. Three-face Franseen needle tip with 15° bevel angle was used in this study.

Figure 5. The MAF polishing experimental setup [11]

3.2. Needle insertion to measure inner friction force and biopsy length

The needle insertion setup to measure the inner friction force in PVC is shown in Fig. 6(a). PVC is a soft tissue-mimicking material for needle insertion experiment because of its similarity to real tissue and the high optical transparency [27]. The ratio of the plastic softener to the original PVC particle solution was 0.5. The plastic softener and the PVC particle solution (M-F Manufacturing Ft. Worth, TX) were mixed with 5 wt% mineral oil (W.S. Dodge Oil Co., Maywood, CA) to simulate the lubrication condition close to that in real tissue [23]. Six PVC column samples (44.5 mm OD) with 10, 20, 30, 40, 50, and 60 mm in length were fabricated. Due to the shrinkage of PVC during curing, the measured lengths of these phantom samples are 9, 18, 24, 33, 42, and 50 mm. In the force measurement setup, the phantom was mounted onto a dynamometer (Model 9256C1, Kistler, Winterthur, Switzerland) with a custom-made fixture.

Two linear stages (Model 200cri, Siskiyou Instrument, Grants Pass, OR) were utilized to drive the needle at 1 mm/s. For each sample, 5 repeated needle insertion tests were conducted.

The experimental setup to measure the biopsy length of PVC phantom was similar to that in the inner friction force measurement (Fig 6(a)). To rotate the phantom, the phantom, dynamometer, and base are rotated 5° around the center of the phantom, as shown in Fig. 6(b). Only the 50 mm PVC phantom was used in this biopsy length experiment. A negative air pressure (-80 kPa) was applied inside the needle in the biopsy length experimental setup to get longer biopsy cores [13].

Figure 6. Biopsy length and friction force measurement setup with PVC samples: (a) picture of the experimental setup and (b) schematic (top view) of the phantom setup.

4. Results

4.1. Inner surface roughness of three types of needles

The surface topographies and surface roughness S_z and S_a of the MAF smoothly polished, roughly polished, and unpolished needles are shown in Figures 7(a) to (c). The area of measurement is $177 \times 132 \mu\text{m}^2$. There are 640×480 ($P \times Q$) points in this area. The whole area was divided into 480 lines in x direction. Each line along the z direction (axial direction of the needle tube) has 640 points. The interval of points is $0.277 \mu\text{m}$ in the z direction and $0.275 \mu\text{m}$ in the x direction.

Figure 7. The typical surface topography of three needle inner surfaces: (a) smoothly polished, (b) roughly polished, and (c) unpolished surfaces.

The smoothly polishing reduced the surface roughness to 0.02 μm Sa . With only two polishing passes, the surface roughness was lowered to 0.26 μm Sa , about 60% of that of the original surface with 0.43 μm Sa . The MAF polishing, generated grooves perpendicular to the sliding direction of the PVC. Although the unpolished surface had the largest surface roughness, its surface topography was more isotropic than those of the MAF polished surfaces.

4.2. The inner and outer friction forces and cutting force of needles

The experimentally measured needle insertion forces are shown in Fig. 8. In the beginning, the insertion force increased with the insertion length until point A. At this moment, the needle tip cut into the PVC and then the needle insertion force dropped to point B. After point B, the insertion force continued to increase with the insertion length. At point P, the insertion force reached the peak F_1 and the needle tip started to punch out of the sample. Points P_s , P_r , and P_u were the penetration point with force F_{1s} , F_{1r} , and F_{1u} for the smoothly polished, roughly polished,

Figure 8. Needle insertion force profiles of three types of needles

The linear fitting of the $F_1 - F_2 (= F_c + F_i)$ vs. length l of the average of 5 repeated insertions is shown in Fig. 9. Values of f_{fi} , f_{fo} , and F_c of three types of needles are summarized in Table 1. As expected, the smoothly polished needle had the smallest f_{fi} (0.0024 N/mm) because the asperities had been removed by MAF and this needle had the lowest inner surface roughness. The roughly polished needle had the largest f_{fi} (0.0041 N/mm) although its inner surface roughness was smaller than that of the unpolished needle.

Values of f_{fo} of the three types of needles were close to each other because of the same polishing condition of the needle outer surface.

The cutting forces F_c of the smoothly and roughly polished needle were close to each other. The unpolished needle had a lower cutting force than those of the other two needles. This was likely due to the variation in cutting edges during the needle tip grinding.

Figure 9. The linear fitting of the relationship between $F_c + F_i$ and l for three types of needles

Table 1. The average f_{fi} , f_{fo} , and F_c of three types of needles

	MAF smoothly polished	MAF roughly polished	Unpolished
f_{fi} (N/mm)	0.0024 ± 0.0010	0.0041 ± 0.0011	0.0034 ± 0.0009
f_{fo} (N/mm)	0.016 ± 0.0012	0.017 ± 0.0009	0.016 ± 0.0015
F_c (N)	0.360 ± 0.033	0.362 ± 0.037	0.331 ± 0.031

4.3. Inner normal stress and coefficient of friction

The inner friction stress σ_{fi} was obtained by dividing f_{fi} with the perimeter of the needle inner surface and listed in Table 2 for three types of needles. Since the material removal amount during MAF was very small, the D_i measured for all three needles were the same (1.08 mm). The average values of D_h created by smoothly polished, roughly polished, and unpolished needles were 1.19, 1.20, and 1.20 mm, respectively. The small difference in the hole diameter was attributed to the inner needle surface roughness. The smoother inner surface resulted in less resistant force for the core to enter the needle and made less core deformation than the rougher needle surfaces. Therefore, the smoothly polished needle had the smallest D_h . The differences between the D_h created by the roughly polished needle and the unpolished needles were too small to be detected. The average normal stress σ_{Ni} obtained from Eq. (5) are also listed in Table 2. The μ , calculated by Eq. (2) and listed in Table 2, was consistent with f_{fi} . The MAF smoothly polished needle had the smallest μ , while the roughly polished needle had the largest μ . The μ of the unpolished needle was between those two types of needles.

Table 2. The average σ_{fi} , σ_{Ni} , and μ of three types of needles

	MAF smoothly polished	MAF roughly polished	Unpolished
σ_{fi} (kPa)	0.70	1.21	1.00
σ_{Ni} (kPa)	2.00	2.16	2.16
μ	0.35	0.56	0.46

4.4. The coefficient of friction calculated from surface topography data based on the ratchet mechanism

The MAF smoothly polished inner needle surface has an Sa about one order lower than the Sa of either of the other two needles. To calculate the coefficient of friction μ_a based on Section 2.3, the measured value μ (0.35) of the smoothly polished needle was assumed to be μ_0 . Table 3 lists μ_a of the smoothly polished, roughly polished, and unpolished inner needle surfaces calculated from the inner needle surface topography data (Fig. 7) and the μ measured from the insertion experiment. The μ_a of the smoothly polished needle was the smallest, while the μ_a of the roughly polished needle was the largest. Moreover, the calculated μ_a values were greater than the measured μ in all three cases. Eq. (8) was derived using μ_0 to denote the coefficient of friction for a perfectly smooth surface. However, the smoothly polished needle surface is not perfectly smooth. Accordingly, μ_a was slightly greater than the measured value μ . For all three types of needle inner surfaces, the calculated μ_a was close to the μ obtained from experiments with less than 15% error. It showed that ratchet mechanism could be used to explain the difference in the friction of the roughly polished and unpolished inner surface.

Table 3. Calculated and measured coefficients of friction μ_a and μ

	MAF smoothly polished	MAF roughly polished	Unpolished
μ_a (calculated using the ratchet mechanism)	0.38	0.58	0.52
μ (measured)	0.35	0.56	0.46

4.5. Biopsy sample length

The average lengths of phantom samples acquired by needle biopsy l_c are summarized in Fig. 10. The needle inner surface with a lower coefficient of friction had a longer biopsy sample length. The l_c obtained with the smoothly polished needle was the longest. The roughly polished needle had the shortest l_c with the largest inner friction force and coefficient of friction. The l_c was longer than 30 mm due to vacuum inside the needle.

Figure 10. Biopsy core length l_c obtained from PVC phantom with three types of needles

5. Discussion

The surface roughly polished using MAF had a larger μ than that of the unpolished surface, and this trend could be attributed to the surface topography produced by MAF. As shown in Fig. 7(c), the unpolished surface shows many asperities randomly distributed on the whole surface. The rough polishing cut off the peaks and reduced the surface roughness. However, the roughly polished surface (Fig. 7(b)) has circumferential abrasive cut marks, which are perpendicular to the direction of PVC flow. When the PVC slides over the needle surface, the micro-ridges superimposed on the irregular asperities might resist the PVC flow, resulting in the larger coefficient of friction than that of the unpolished surface. The surface roughness S_a of the smoothly polished needle was about one order of magnitude lower than the S_a of either of the other two. The smoothly polished surface (Fig. 7(a)) consists of an accumulation of periodical cut marks in the circumferential direction of the needle. However, the surface is much smoother than the other two, the coefficient of friction of the smoothly polished needle was the lowest of the three.

The ratchet mechanism could be used to discuss the surface topography effect on the friction. The calculation results based on the ratchet mechanism showed the same trend of the μ_a of the three needle inner surfaces, which supports the surface topography explanation. The surface roughness S_a was not the only factor to determine the coefficient of friction between PVC and needle inner surface.

The biopsy length is affected by the inner friction force of the needle. The smaller the inner friction force is, the longer the biopsy length is. Reducing the inner friction force could lower the

resistance to the phantom core to move inside the needle and increase the biopsy length. In this paper, the MAF smoothly polished needle had the longest biopsy length because it possessed the lowest inner friction force among the three types of needles.

6. Conclusions

The inner friction force and biopsy length of MAF smoothly polished, roughly polished, and unpolished needle were measured and compared in PVC phantom material. The smoothly polished needle had the smallest f_i (0.0024 N/mm) and longest l_c (34.3 mm) while the roughly polished needle had the largest f_i (0.0041 N/mm) and shortest l_c (30.5 mm). The normal stress at the inner surface was calculated with the deformation of the phantom core to obtain the coefficient of friction μ . Similar to f_i , the μ of the smoothly polished needle was the smallest (0.35) among three needles. The roughly polished needle had the largest μ (0.56). The existence of circumferential ridges and grooves on the surface topography of the roughly polished needle created large resistance to the biopsy core movement and made the f_i and μ of the roughly polished needle larger than those of the unpolished needle. The apparent coefficient of friction calculated from the coordinates of the surface topography was closed to the measured μ , which showed that the difference between the three types of surfaces was caused by the surface topography. Based on the results of the biopsy length obtained from PVC samples, larger f_i and μ led to smaller biopsy length. Smoothly polishing the needle inner surface with MAF could effectively lower the f_i and increase the l_c .

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