Friction **8**(1): 58–69 (2020)
https://doi.org/10.1007/s40544-018-0242-6
ISSN 2223-7690
CN 10-1237/TH
RESEARCH ARTICLE

Dopamine hydrochloride and carboxymethyl chitosan coatings for multifilament surgical suture and their influence on friction during sliding contact with skin substitute

Gangqiang ZHANG[1,2,3,*], Ganlin ZHENG[3], Tianhui REN[3], Xiangqiong ZENG[1,4,*], Emile van der HEIDE[4,5,6]

[1] *Advanced Lubricating Materials Laboratory, Shanghai Advanced Research Institute, Chinese Academy of Sciences, Shanghai 201210, China*

[2] *Institute of Functional Textiles and Advanced Materials, Colloge of Textlies & Clothing, Qingdao University, Qingdao 266000, China*

[3] *School of Chemistry and Chemical Engineering, Key Laboratory of Thin Film and Microfabrication Technology (Ministry of Education), Shanghai Jiao Tong University, Shanghai 200240, China*

[4] *Laboratory for Surface Technology and Tribology, University of Twente, Enschede 7500AE, The Netherlands*

[5] *TU Delft, Faculty of Civil Engineering and Geosciences, Stevinweg 1, Delft 2628 CN, The Netherlands*

[6] *Soete Laboratory, Ghent University, Zwijnaarde B-9052, Belgium*

Received: 19 March 2018 / Revised: 25 July 2018 / Accepted: 12 August 2018

Abstract: In order to reduce the damage to tissue and fill the interstices between fibers, multifilament sutures are frequently treated with certain coating materials. The objective of this study was to create and characterize dopamine hydrochloride (DA) and carboxymethyl chitosan (CMCS) coatings on surgical sutures and investigate their effects on the frictional performance of the surgical sutures during sliding through a skin substitute. The effects of the treatment on the physical and chemical characteristics of the surgical sutures were evaluated. The friction force of the surgical sutures during sliding through the skin substitute was experimentally determined using a penetration friction apparatus. The coefficient of friction (COF) was calculated using a linear elastic model and was used to estimate the frictional behavior of the surgical suture-skin interactions.

The results showed that the DA coating could evenly deposit on the surface of the etched multifilament surgical suture surfaces in a weakly alkaline buffer solution. The CMCS coating material could form a uniform film on the surface of the sutures. Minor changes in the surface roughness of the multifilament surgical sutures with different treatments occurred in this study. The friction force and the COF of the multifilament surgical sutures with DA and CMCS coating showed little change when compared with untreated multifilament surgical sutures.

Keywords: friction; surgical suture; coating; Biomaterials; skin substitute

1 Introduction

Surgical sutures are available in different sizes and structures and might be coated or uncoated depending on the surgical procedure they are used for. The frictional behavior of surgical sutures plays an important role in their interaction with tissue because it determines how easily a suture passes through tissue and is related to the amount of tissue trauma that they cause.

The frictional performance of fibrous materials is largely governed by fiber-related factors, such as geometrical structure, surface characteristics, and materials used [1, 2]. When a surgical suture slides

* Corresponding authors: Gangqiang ZHANG, E-mail: cris8866@163.com; Xiangqiong ZENG, E-mail: zengxq@sari.ac.cn

through tissue with high friction, it damages tissue, leading to longer recovery times for wounds [3]. For soft or fragile tissue, uncontrolled stitch cracking may cause additional suturing-related trauma.

In general, surgical sutures are categorized as either monofilament or multifilament according to their structure [4, 5]. Monofilament and multifilament surgical sutures have different friction behaviors during surgery. Multifilament sutures tend to have better mechanical properties with a significantly higher flexibility and pliability than monofilament sutures. However, multifilament surgical sutures with a twisted structure have higher surface roughness [6], which in turn can influence their frictional behavior.

For uncoated multifilament sutures, a twisted structure and rough surfaces may lead to increased resistance when the suture slides through tissue. The interstices between fibers could harbor bacteria within the braided structure [7]. In order to reduce the damage caused to tissue and fill the interstices between fibers, multifilament sutures are frequently treated with some coating materials.

Researchers have investigated the influence of various coating materials on multifilament sutures, such as triclosan [8], chitosan [9, 10], silver nanoparticles [11], and antimicrobial peptides [12, 13], on the physical performance of multifilament surgical sutures. However, the main focus of these investigations was to reduce tissue reactions. The influence of coating materials on the frictional behavior of surgical suture-skin interactions has rarely been investigated. Walling et al. [14] studied the influence of antibiotic ointments on the friction behavior of polyglactin surgical sutures. Their results showed that the coefficient of friction of sutures decreased and the suture easily passed through tissue when using the lubricant. Chen et al. [15] investigated the friction force of coated antibacterial braided silk interacting with a skin substitute. They found that the maximum friction force decreased when using the coating material.

Coating materials should be biocompatible with tissues and not provoke chronic tissue reactions. Dopamine hydrochloride (DA) [16] and carboxymethyl chitosan (CMCS) [9] are widely used in various biomedical applications and were chosen as coating materials in this study. Recently, DA, which is a novel material, has been used to modify surfaces to make them cohesive and functionalized [16, 17]; this has drawn widespread attention. DA coating was found to be a versatile platform for secondary reactions [18]. In a buffer solution, DA can be self-polymerized and deposited on hydrophobic or hydrophilic surfaces [19]. Carboxymethyl chitosan (CMCS) with the CH_2-COOH group, which is derived from chitosan, has been used as a coating and biomedical material [20]. Viju et al. [21] observed that ,compared with a chitosan-untreated silk suture, the chitosan-treated sample presented a lower coefficient of friction between fibers for normal loads ranging from 50 g to 150 g. To the best of our knowledge, there has been hardly any research on the friction of surgical sutures with combined DA and CMCS coating.

The objective of this study was to apply DA and CMCS coatings on multifilament surgical sutures and investigate the influence of the DA and CMCS coating on their frictional performance when sliding through a skin substitute. The treating process for multifilament sutures was optimized. The effects of the treatment process on the physical and chemical characteristics of the suture were evaluated with different parameters, including weight change, contact angle, tensile strength, stiffness, laser surface characteristics, and surface chemical composition. The frictional performance of the coated sutures was studied using the recently developed penetration friction apparatus (PFA) [22]. The coefficient of friction based on the linear elastic model was used to estimate the frictional behavior of surgical suture-skin interactions.

2 Materials and methods

2.1 Materials

Polyglycolic acid (PGA) multifilament surgical sutures and straight stainless-steel tapered needles with a chamber at the end (Weigao Medical Instruments Co. Ltd., Weihai, China) were used in this study. The skin substitute was Sil8800 (Red, 80 IRHD), obtained from Superior Seals, with a similar toughness and constitution as skin [23–25], as shown in Fig. 1. Dopamine hydrochloride (DA) and carboxylated chitosan (CMCS) were obtained from Aladdin Chemistry

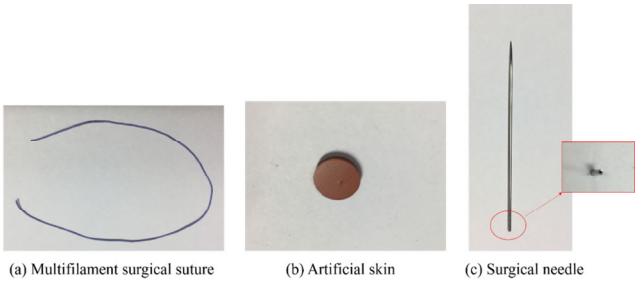

Fig. 1 Multifilament surgical suture, artificial skin, and surgical needle.

(Shanghai, China). Tetrahydrofuran (THF) and sodium hydroxide (NaOH) were purchased from Sigma-Aldrich (St. Louis, USA).

2.2 Coating process

The coating process of the PGA multifilament surgical sutures involved the following three steps. First, in order to improve the adhesion of the coating material to the PGA multifilament surgical suture, the latter was cleaned with a THF solution to remove the protective layer and etched using NaOH solution. Secondly, the etched PGA multifilament surgical suture was coated with DA *via* an immersed coating method. Finally, the resulting DA-coated PGA multifilament surgical suture was coated with CMCS *via* a dip coating method. The coating process is illustrated in Fig. 2.

2.2.1 *Cleaning of the PGA surgical suture*

The PGA multifilament surgical suture samples (~25 cm long) were immersed in THF (20 mL) and subjected to 5 min of ultrasonic treatment. Then, the samples were treated with fresh boiling THF (50 mL) under 12 h of reflux. The suture samples were washed with fresh THF and dried at room temperature.

2.2.2 *Etching of the cleaned PGA sutures*

The cleaned PGA multifilament surgical sutures were etched using NaOH solution with a concentration of 0.50 mol/L at room temperature. The PGA multifilament surgical sutures, which had a length of 25 cm, were immersed in a beaker containing the NaOH solution and subjected to magnetic stirring over 1 min, 2 min, 3 min, 4 min, and 5 min (respectively for each sample). Then, the PGA multifilament surgical sutures were washed with deionized water and dried at room temperature. The optimization of the etching time for PGA multifilament surgical sutures was investigated.

2.2.3 *Dopamine coating*

DA powder was dissolved in 10-mM Tris buffer. The effects of pH value (7.5, 8.5, and 9.5) and the concentration of the DA solution (1 mg/ml–4 mg/ml) on the coating were investigated. The etched PGA multifilament surgical suture samples were immersed into the DA solution and stirred at room temperature for 12 h. Then, the PGA multifilament surgical sutures were washed with deionized water. After being dried to a constant weight at 40 ℃, the resulting modified sutures were used for characterization.

2.2.4 *CMCS coating*

The CMCS coating slurry was prepared by slowly dissolving CMCS in deionized water *via* magnetic stirring. The solution, which had a concentration of 10 mg/mL, was sonicated for 5 min. The PGA multifilament surgical sutures were treated *via* a dip-coating method. The PGA multifilament surgical sutures were dipped into the CMCS slurry once or several times, leading to the formation of a homogeneous coating layer on the surface of the sutures. The PGA multifilament surgical sutures with DA coating were immersed in the CMCS solution for 30 s and then removed. The PGA multifilament surgical sutures were pulled tight and tilted. The coating solution droplets were moved back and forth on the sutures until they visibly disappeared. This step was repeated according to the test design. Finally, the sutures were dried at room temperature for 30 min.

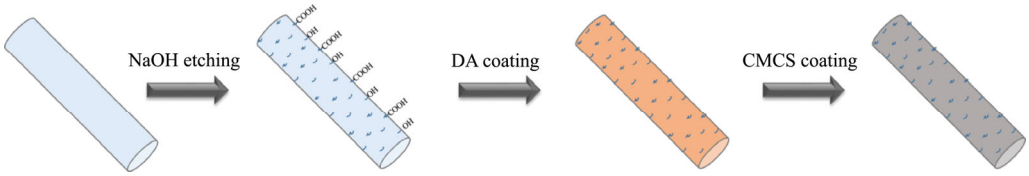

Fig. 2 Schematic illustration of the coating process of the PGA multifilament surgical suture.

2.3 Tribological measurements

2.3.1 Sample preparation

Before our frictional experiments, the PGA multifilament surgical sutures were cut into samples that were 25 cm long. Circular-shaped skin substitute samples were prepared with a diameter of 23 mm and a thickness of 2 mm. The samples were cleaned in acetone under ultrasonic treatment for 10 min and then dried before the experiment.

2.3.2 Analysis methods

2.3.2.1 Weight change

The weight change (w) of the etched and coated PGA multifilament surgical sutures was calculated using the weight of an untreated suture (w_1) and the weight of a treated one (w_2) as shown in Eq. (1).

$$w = \frac{w_2 - w_1}{w_1} \times 100\% \tag{1}$$

Weight change was positive for the etched PGA multifilament surgical sutures and negative for the coated PGA multifilament surgical sutures.

2.3.2.2 Contact angle

The surface hydrophilicity of the sutures was characterized *via* water contact angle (CA) measurements (CA, Data Physics OCA20, Germany) using the sessile drop method at room temperature [26]. In each measurement, a tight PGA multifilament surgical suture was fixed on the tester, and a 3-μL drop of water was placed on it in air. A video camera recorded images of the water droplet and measured the contact angle. Software (OCA20 software, Data Physics Instruments, Germany) was used to determine the contact angles from the images. Five measurements were performed for each PGA multifilament surgical suture, and the mean values were taken as the final results.

2.3.2.3 Tensile strength tests

The influence of the treatment on the strength of the PGA surgical sutures was measured using a tensile tester (Zwick/Roell 500N, Germany). In this experiment, one end of the suture (with a length of 10 cm) was clamped in a fixed jaw, whereas the other end was fixed to the moving jaw of the tensile tester. With the velocity of the moving jaw set to 10 mm/min, the suture was pulled until it broke. The computer recorded the maximum load [27].

2.3.2.4 Bending stiffness

The bending stiffness of the PGA multifilament surgical sutures affects the shear strength of the sliding friction, thereby affecting the frictional behavior of the surgical suture-skin substitute interactions [28]. The bending stiffness of the PGA multifilament surgical sutures was evaluated using the cantilever beam law. In each test, 5 cm of the PGA multifilament surgical suture were placed on a horizontal plane. Afterward, 1 cm of the PGA multifilament surgical suture was left to hang in the air. Then, a dead load was attached to this end. The end of the PGA multifilament surgical suture fell down a certain distance, which was measured after 15 s of loading [29]. The bending stiffness was then calculated according to Eq. (2),

$$B = \frac{F * l^3}{3f} \tag{2}$$

where B is the bending stiffness, l was 1 cm in this study, and f is the dead load applied to the end of the PGA multifilament surgical suture.

2.3.2.5 Surface analysis of materials

The topography and 3D profiles of the PGA multifilament surgical sutures were obtained using a VK-9700 laser scanning microscope from KEYENCE, Japan. Characterization of the PGA multifilament surgical sutures was carried out using a spectrum 100 FT-IR spectrophotometer (Perkin Elmer) over a wave number range from 600 cm^{-1} to 4,000 cm^{-1}.

2.3.2.6 Experiment setup

The frictional performance of the PGA multifilament surgical sutures for a tissue substitute was investigated using the recently developed PFA [22]. The main parts of the PFA are briefly described in this section.

The main component of the PFA is the sample gripper, which is used to fix the tissue substitute sample in place. It was assembled inside the Zwick/Roell tensile tester (500 N, Germany) to measure the friction force

while the PGA multifilament surgical suture penetrated through the skin substitute. In the tribological measurements, the skin substitute was attached to the gripper (Fig. 3(a)). In addition, the PGA multifilament surgical suture penetrated through the skin substitute with a surgical needle and was attached to a load cell. The load cell recorded the friction force between the PGA multifilament surgical suture and the skin substitute for the movements of the sample gripper (Fig. 3(b)). Table 1 shows an overview of the experimental conditions. Three repeated experiments were carried out for each set of experimental conditions.

3 Results and discussion

3.1 Process optimization

3.1.1 *Etching of the PGA sutures*

A smaller contact angle (CA) on the surface means that the fabric has better hydrophilicity. The CA of the PGA multifilament surgical sutures before and after etching were measured and are shown in Fig. 4. It can be seen that the water CA decreased as the NaOH treating time increased. Compared with the untreated PGA multifilament surgical sutures, the NaOH treated

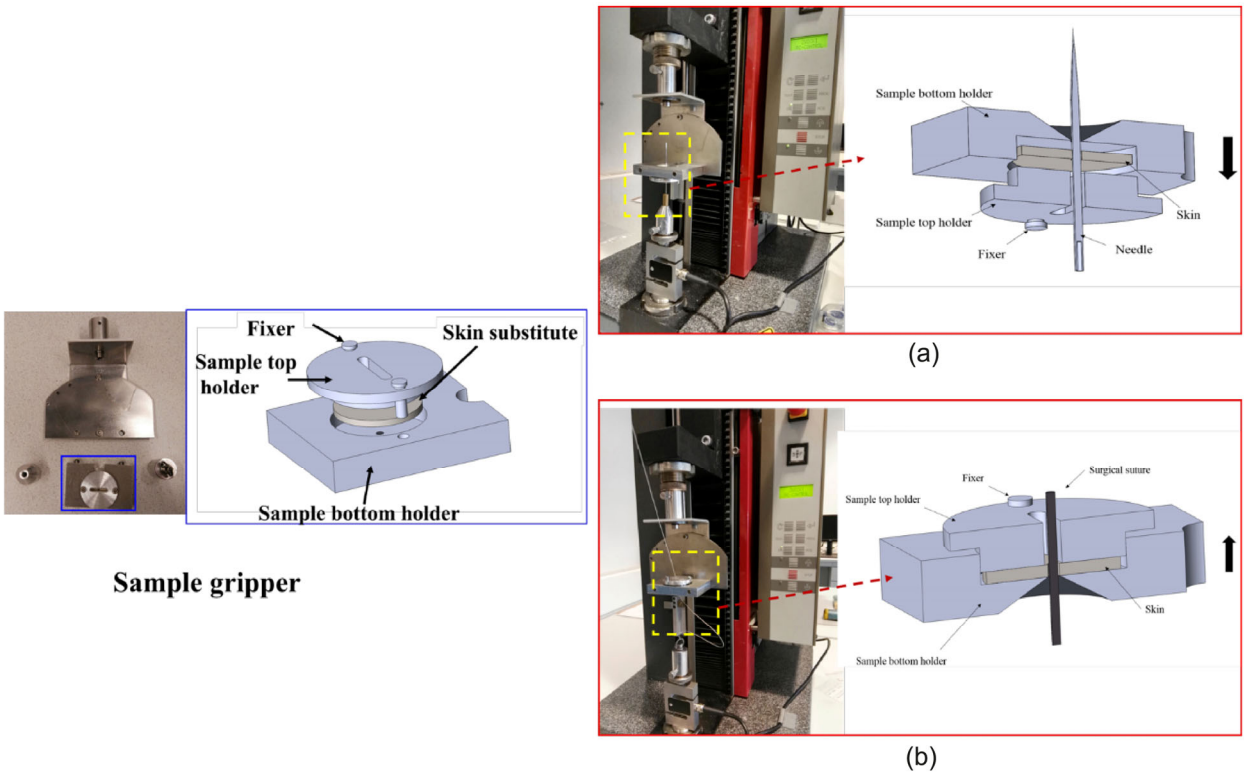

Fig. 3 Penetration friction apparatus (PFA). (a) punch process; (b) penetration process.

Table 1 Experimental parameters.

Test–related condition	Specification
Equipment	Penetration friction apparatus
Diameter of space of gripper	25 ± 0.5 mm
Puncture angle	90°
Puncture velocity of needle	60 mm/min
Puncture distance of needle	10 mm
Penetration velocity of suture	100 mm/min
Penetration distance of suture	100 mm

sutures had a smaller CA. That is because alkaline hydrolysis reactions increased the number of -OH and -COOH groups. When the PGA multifilament surgical sutures were treated with NaOH for 5 min, their CA decreased from 66.1° to 55.4°, subsequently increasing the hydrophilicity of the sutures.

The weight changes of the PGA multifilament surgical sutures for different etching parameters were considered in this study. Figure 5 shows the weight changes of the PGA multifilament surgical suture etched using a 0.05-mol/L NaOH solution for various treating times. It can be seen that the weight loss of the sutures increased with etching time. We also noted that the weight loss was 16.5% for NaOH treatment of 5 min.

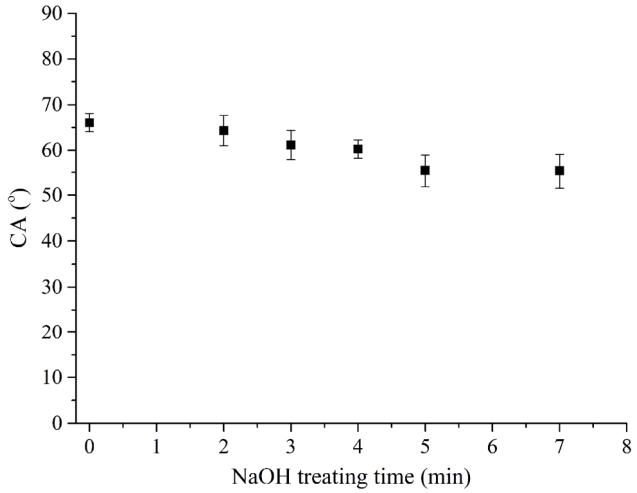

Fig. 4 Water contact angle of the PGA multifilament surgical sutures as a function of NaOH treating time.

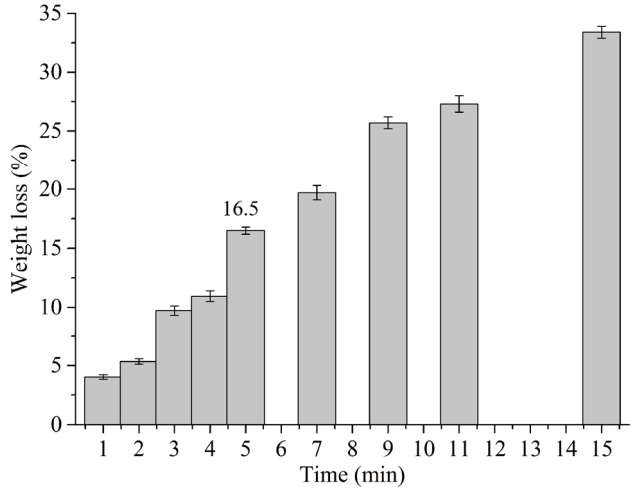

Fig. 5 Weight loss of PGA surgical sutures treated with NaOH.

When choosing a suture, its strength must be considered. A suitable suture strength will adequately hold the healing tissue. The effects of etching treatments on the tensile properties of the PGA multifilament surgical sutures are shown in Fig. 6. It can be observed from the curve that the tensile force and the tensile elongation decreased as the NaOH treating time increased. That is because the NaOH altered the chemical structure of the suture and affected its physical performance. Treatment of the PGA multifilament surgical sutures with NaOH for 5 min resulted in a slight reduction in tensile strength from 58.9 N to 50.0 N and tensile elongation from 30.4% to 24.2%.

We therefore concluded that the NaOH etchant is efficient for PGA suture surface treatment. Based on the above analysis, the NaOH etching conditions were set as follows: NaOH concentration of 0.05 mol/L, room temperature, and etching time of 5 min.

3.1.2 Coating

3.1.2.1 DA coating

In these experiments, DA was dissolved in Tris buffer solutions with different pH values (10 mM; pH 7.0, 8.5, and 9.5) with a concentration of 3 g/L. Figure 7 shows that the coloring of the DA solution increased as pH value increased. That is because DA aggregates at higher pH values. The color of the DA solution showed gradual change from colourless to dark brown [30]. However, when the pH value was 9.5, the DA aggregated into larger groups, leading to an uneven DA-coating film on the surface of the PGA multifilament surgical suture.

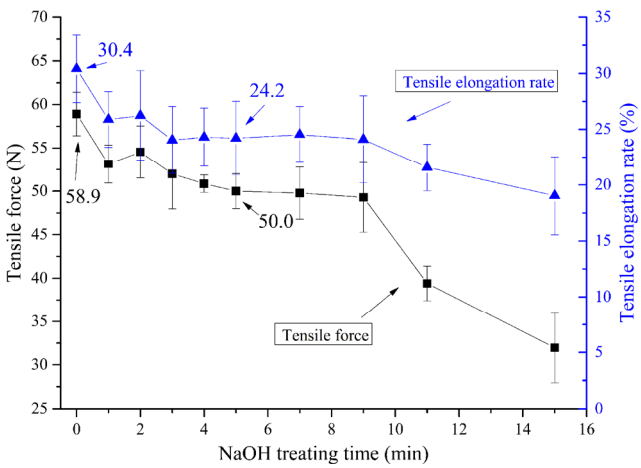

Fig. 6 Tensile force and elongation rate of the PGA surgical sutures after NaOH treatment.

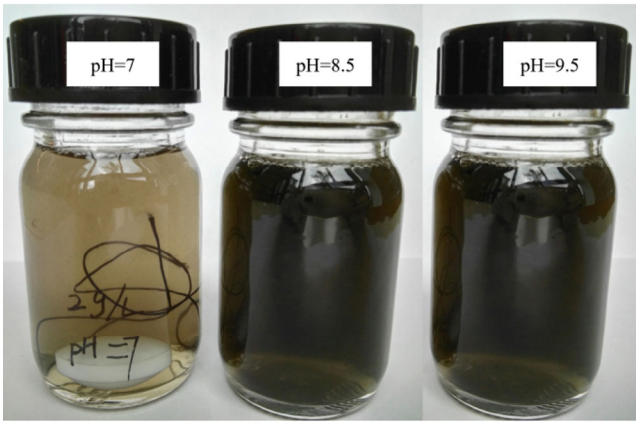

Fig. 7 Digital picture of DA solutions with different pH values.

The effects of DA concentration on deposition was investigated for various concentrations (1 g/L, 2 g/L, 3 g/L, and 4 g/L) at a pH of 8.5. Figure 8 shows the concentration of DA versus the weight of the PGA multifilament surgical suture. It can be seen that the weight of surgical sutures increased from 2.75% to 4.60% as concentration increased from 1 g/L to 4 g/L.

The morphologies of the coated PGA multifilament surgical sutures are shown in Fig. 9. It can be seen that DA deposition increased as the concentration of DA increased. Furthermore, when the DA concentration increased to 3 g/L, a further increase in the concentration did not significantly increase deposition. Hence, the optimum conditions for DA coating were determined to be as follows: a pH value of 8.5 and a concentration of 3 g/L.

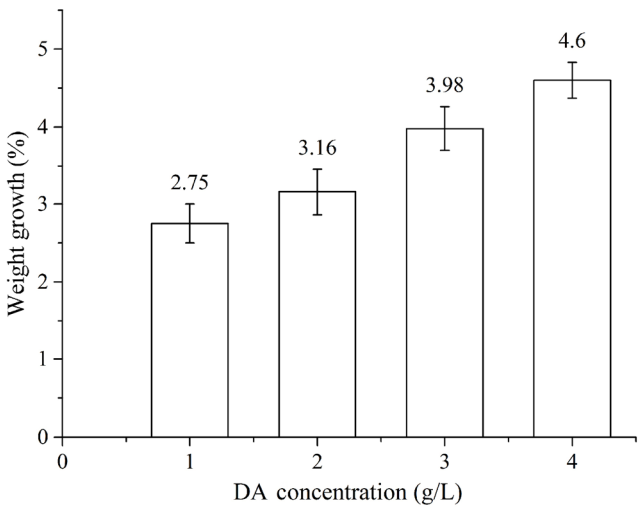

Fig. 8 Weight growth of the PGA multifilament surgical suture for different DA concentrations.

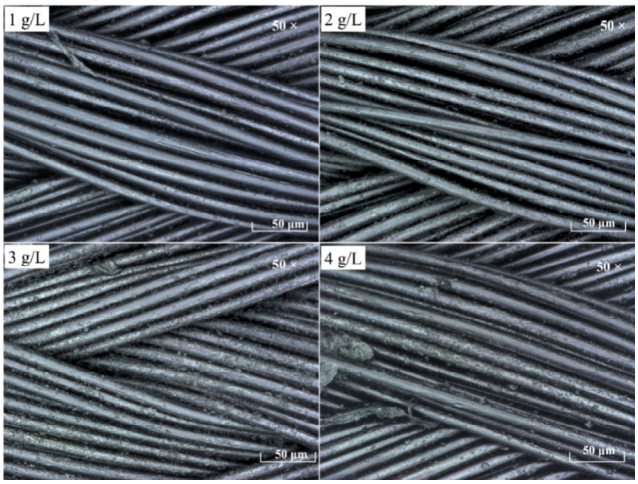

Fig. 9 Morphologies of the surfaces of the PGA multifilament surgical sutures for different DA concentrations ranging from 1 g/L to 4 g/L at a pH of 8.5 (magnification 50×; one pixel represents 0.135 mm × 0.135 mm).

3.1.2.2 CMCS coating

Figure 10 shows the weight growth of the PGA multifilament surgical sutures after coating with CMCS. It can be seen that the weight of the sutures after being dipping into the coating solution increased after repeating the coating process multiple times. After coating three times, the weight increase rate of the surgical sutures lowered. Therefore, we decided to coat the sutures three times in this study.

3.2 Characterization

3.2.1 NaOH etching

The PGA sutures were treated with NaOH to generate carboxyl groups on the surface of the PGA multifilament surgical sutures. Both the untreated and treated PGA multifilament surgical sutures were characterized *via* FT-IR spectroscopy to confirm the occurrence of this chemical reaction. The FT-IR spectra of the untreated and treated PGA multifilament surgical sutures in the region from 4,000 cm^{-1} to 700 cm^{-1} are presented in Fig. 11. The significant features of the carboxyl groups are the bands corresponding to the O-H stretching of CO-OH at 3,500 cm^{-1}, the bending vibrations that appeared at 1,263 cm^{-1} and 1,415 cm^{-1}, and the C=O carbonyl stretching of -COOH that appeared at 1,737 cm^{-1}. It can be seen that the spectrum of treated PGA multifilament surgical suture was similar to that of the unetched PGA multifilament

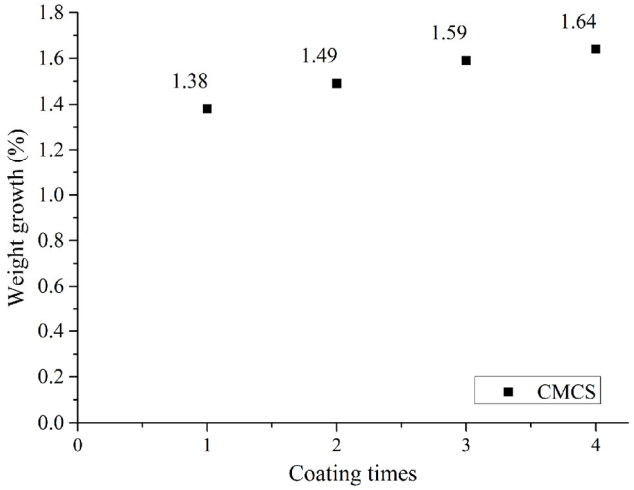

Fig. 10 Weight growth of the PGA multifilament surgical sutures versus times coated with CMCS.

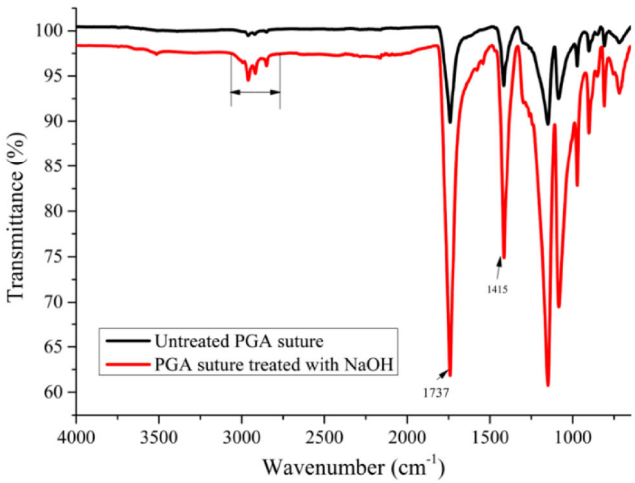

Fig. 11 FT-IR spectra of the PGA surgical sutures untreated and treated with NaOH.

surgical suture, but a stronger absorption band of carboxymethylation stretching was observed on the surface of the etched PGA multifilament surgical suture.

The morphologies of the untreated and treated PGA multifilament surgical sutures were observed using a laser scanning microscope at the voltage and magnification conditions shown in Fig. 12. We found that, after cleaning, the surface of the suture became smooth. After NaOH etching, some dents appeared on the surface of the surgical sutures.

3.2.2 *Coating*

3.2.2.1 *FT-IR analysis*

The chemical structures of the coated PGA multifilament surgical sutures were also confirmed *via* FT-IR measurements, as shown in Fig. 13. Some changes in the FT-IR spectra of the PGA multifilament surgical sutures were observed after coating with DA and CMCS. Compared with the uncoated PGA multifilament surgical sutures, several new absorption signals appeared after surface coating. The absorbance of the stretching vibrations of N-H/O-H was observed between 3,600 cm^{-1} and 3,100 cm^{-1}. The overlap of the resonance vibrations of the C=C bonds in the aromatic ring corresponded to the peak at 1,400 cm^{-1}, and the

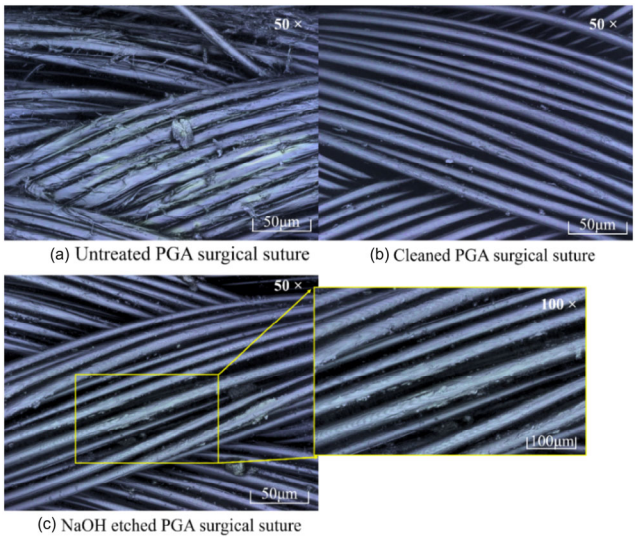

Fig. 12 Morphologies of the untreated and treated PGA multifilament surgical sutures. (a) untreated PGA surgical suture; (b) cleaned PGA surgical suture; (c) NaOH-etched PGA surgical suture.

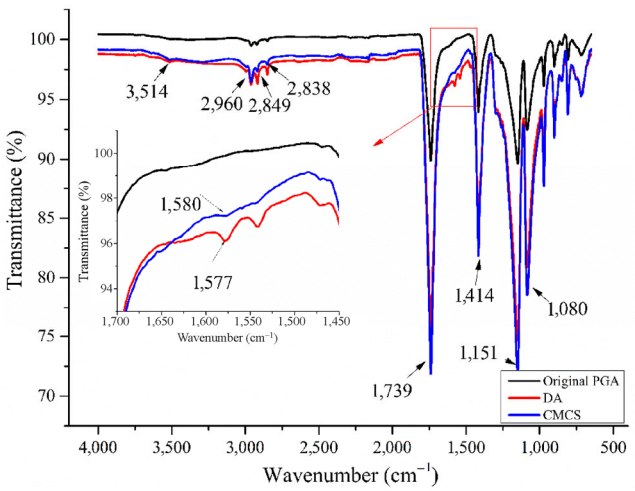

Fig. 13 FT-IR spectra of PGA surgical sutures with different coatings.

N-H bending vibrations corresponded to the peak at 1,700 cm^{-1}. Meanwhile, the absorption peaks at 3,514 cm^{-1} (O-H stretching), 2,838 cm^{-1} (C-H stretching), 1,656 cm^{-1} (amide band), 1,577 cm^{-1} (N-H bending), 1,151 cm^{-1} (O-bridgestretching), and 1,082 cm^{-1} (C-O stretching) corresponded to the basic characteristics of CMCS [31–33]. The characteristics of chitosan were observed from the peaks at 1,730 cm^{-1} (CO-OH), 1,080 cm^{-1}–1,155 cm^{-1} (C-O), and 1,626 cm^{-1} and 1,518 cm^{-1} (NH_3^+) [33]. These results proved the incorporation of the DA and CMCS composite layers on the surface of the PGA multifilament surgical sutures.

3.2.2.2 *Bending stiffness of the surgical sutures*

Bending stiffness is an important factor that affects the frictional behavior and handling characteristics of PGA multifilament surgical sutures. Figure 14 shows the bending stiffness of PGA multifilament surgical sutures with different treatments. It can be seen that the bending stiffness of the original suture was 0.026 $cN^2 \cdot cm^2$, which was higher than that (0.007 $cN^2 \cdot cm^2$) of the suture treated with NaOH. After DA coating, the bending stiffness of the suture slightly increased from 0.007 $cN^2 \cdot cm^2$ to 0.008 $cN^2 \cdot cm^2$. Then, after further coating with CMCS, the bending stiffness of the suture increased to 0.025 $cN^2 \cdot cm^2$, which was similar to that of the original suture. A similar result indicating that the stiffness of sutures increases after coating was found in a previous report [34].

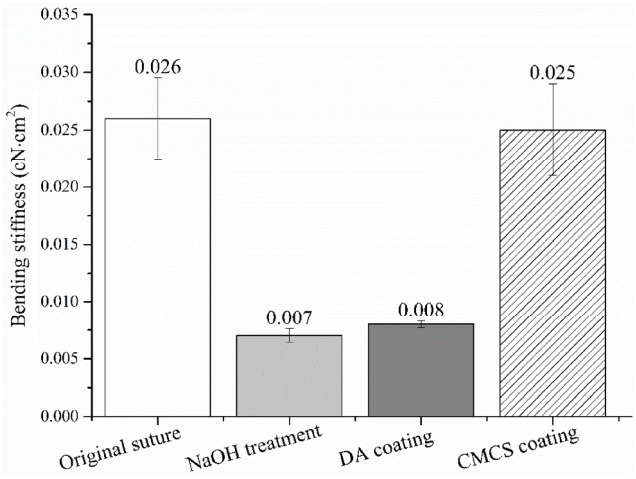

Fig. 14 Bending stiffness of PGA multifilament surgical suture with different treatments.

3.2.2.3 *Surface morphology*

Laser confocal images of the PGA multifilament surgical sutures coated with DA and then coated with CMCS are shown in Fig. 15. It can be seen that the coating materials formed a uniform film on the surface of the sutures. After comparing DA and CMCS coating, we found that the PGA multifilament surgical suture with CMCS coating had become more uniform. Meanwhile, the surface roughness of the PGA multifilament surgical suture with a different treatment at 50× magnification is shown in Table 2. It can be seen that the surface roughness values before and after treatment were very similar.

3.3 Frictional performance

Figure 16 shows the friction force arising from the interactions of the PGA multifilament surgical sutures with different treatments with the skin substitute. It can be seen that when the original suture was cleaned using THF, the friction force decreased from 0.75 N to 0.72 N. When the PGA multifilament surgical suture was treated with DA coating, the friction force increased to 0.75 N. Subsequently, the friction force of the PGA multifilament surgical suture treated with CMCS coating increased to 0.80 N.

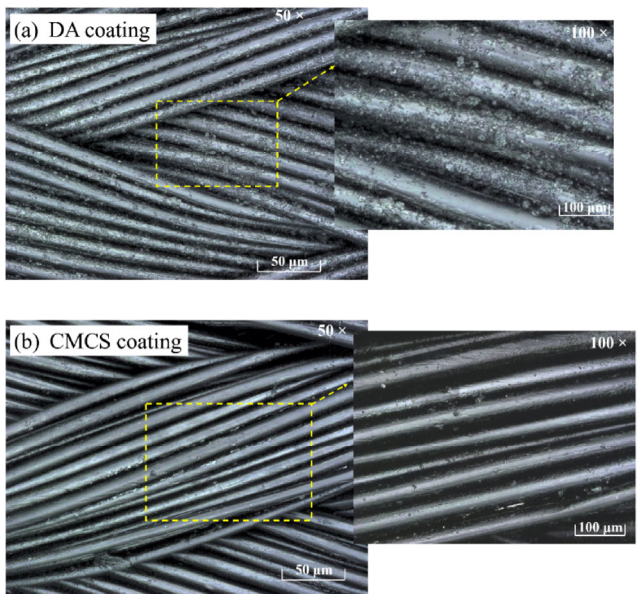

Fig. 15 Morphology of the PGA multifilament surgical sutures. (a) surgical suture with DA CMCS coating; (b) surgical suture with CMCS coating.

Table 2 Roughness of the PGA surgical sutures with different treatments (50× magnification).

Roughness	Original suture	THF treated	NaOH etching	DA coating	CMCS coating
S_a	15.6 ± 0.45	14.3 ± 0.67	15 ± 0.74	15.8 ± 0.67	16.2 ± 0.58
S_q	20.1 ± 0.36	21.7 ± 0.56	19.6 ± 0.54	18.8 ± 0.83	18.8 ± 0.37

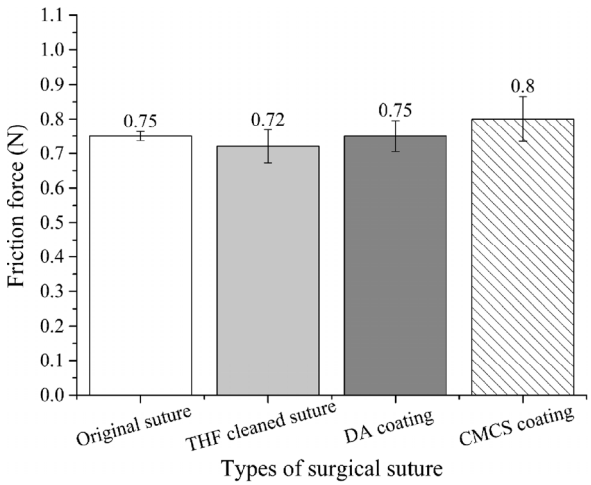

Fig. 16 Friction force of the PGA multifilament surgical sutures with DA and CMCS coating when sliding.

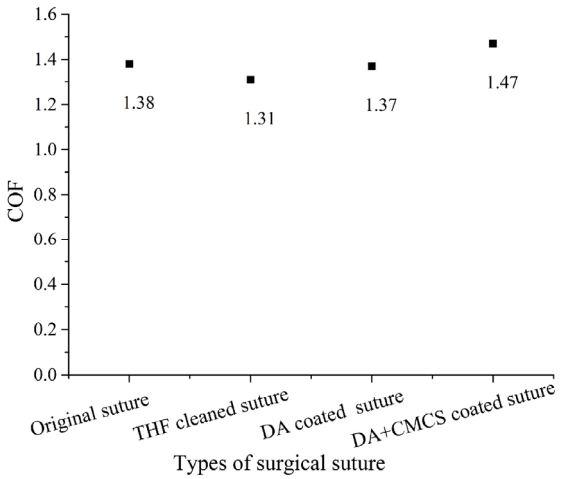

Fig. 17 Coefficient of friction (COF) of the PGA multifilament surgical sutures with different coatings.

In this study, the diameter of the PGA multifilament surgical sutures was consistent before and after treating. Based on the contact model between the surgical suture and skin substitute, a simplified linear elastic model was developed to predict the coefficient of friction. According to this linear elastic model, the normal force was 0.55 N when the suture was sliding through the skin substitute while following the surgical needle. Figure 17 shows the coefficient of friction (COF) of the PGA multifilament surgical sutures with different treatments. It can be seen that the coefficient of friction of the original surgical suture was 1.38. When the PGA multifilament surgical suture was treated with THF, the coefficient of friction decreased to 1.309. After coating with DA and CMCS, the coefficient of friction increased to 1.37 and 1.47, respectively. The latter values are within the range of those of the uncoated suture. The trend of the COF shown in Fig. 17 is the same as that in Fig. 16.

These results may be explained by minimal changes to the high stiffness and roughness of the original suture after coating. This suggests that it is necessary to decrease the surface roughness and increase the flexibility of coated surgical suture, which should be explored in further research. In addition, one of the experimental conditions used in this study was dry friction. In real applications on human tissue, body fluids could act as lubricants, the hydrophilic polymeric coatings might reduce the frictional performance of the suture-tissue interactions.

4 Conclusion

In this study, DA and CMCS coating on PGA multifilament surgical sutures was performed. The PGA multifilament surgical sutures were etched with a NaOH solution. The etched PGA multifilament surgical sutures were coated with DA *via* an immersed coating method. Finally, the resulting DA-coated PGA multifilament surgical sutures were coated with CMCS *via* a dip coating method.

The optimum conditions for etching with NaOH were as follows: NaOH concentration of 0.05 mol/L, room temperature, and an etching time of 5 min. After the PGA suture was etched using NaOH, carboxyl groups appeared on the surface of the PGA sutures, leading to an enhancement of their hydrophilicity.

The DA coating layer could be deposited well on the surface of the PGA multifilament surgical sutures using a weakly alkaline buffer solution. The optimum

conditions for DA coating were as follows: concentration of 3 g/L and pH value of 8.5. The CMSC coating materials could form a uniform film on the surface of the sutures. DA and CMCS coating had barely any effect on the roughness and topography of the PGA multifilament surgical sutures.

The COFs of the PGA multifilament surgical sutures with DA and CMCS coating were 1.37 and 1.47, respectively, which was within the range of that of the uncoated PGA multifilament surgical sutures. The same result was found from friction force data.

Acknowledgments

The authors are grateful to Marie Curie CIG (Grant no. PCIG10-GA-2011-303922), the Shanghai Natural Science Foundation (Grant no. 17ZR1442100), and the Shanghai Municipal "Science and Technology Innovation Action Plan" International Cooperation Project (no. 15540723600) for the financial support.

References

[1] Raes G, Röder H L. The relation between fibre friction and the behaviour of fibres and yarns during processing. *J Text Inst Proc* **46**(1): P84–P100 (1955)

[2] Zhang G Q, Ren T H, Zeng X Q, Van Der Heide E. Influence of surgical suture properties on the tribological interactions with artificial skin by a capstan experiment approach. *Friction* **5**(1): 87–98 (2017)

[3] Apt L, Henrick A. "Tissue-drag" with polyglycolic acid (Dexon) and polyglactin 910 (Vicryl) sutures in strabismus surgery. *J Pediatr Ophthalmol* **13**(6): 360–364 (1976)

[4] Gilbert J L. Wound closure biomaterials and devices. *Shock* **11**(3): 226 (1999)

[5] Moy R L, Waldman B, Hein D W. A review of sutures and suturing techniques. *J Dermatol Surg Oncol* **18**(9): 785–795 (1992)

[6] Rao Y Q, Farris R J. A modeling and experimental study of the influence of twist on the mechanical properties of high-performance fiber yarns. *J Appl Polym Sci* **77**(9): 1938–1949 (2000)

[7] Ajmeri J R, Ajmeri C J. Surgical sutures: the largest textile implant material. In *Medical Textiles and Biomaterials for Healthcare*. Anand S C, Kennedy J F, Miraftab M, Rajendran S, Eds. Boca Raton: Woodhead Publishing, 2006: 432–440.

[8] Storch M, Scalzo H, Van Lue S, Jacinto G. Physical and functional comparison of Coated VICRYL* Plus Antibacterial Suture (coated polyglactin 910 suture with triclosan) with Coated VICRYL* Suture (coated polyglactin 910 suture). *Surg Infect* **3 Suppl 1**: S65–S77 (2002)

[9] Lv J C, Zhou Q Q, Liu G L, Gao D W, Wang C X. Preparation and properties of polyester fabrics grafted with O-carboxymethyl chitosan. *Carbohydr Polym* **113**: 344–352 (2014)

[10] Hernandez-Richter T, Schardey H, Löhlein F, Heiss M, Redondo-Müller M, Hammer C, Schildberg F W. The prevention and treatment of vascular graft infection with a triclosan (Irgasan®)-bonded Dacron graft: an experimental study in the pig. *Eur J Vasc Endovasc Surg* **20**(5): 413–418 (2000)

[11] Zhang S W, Liu X L, Wang H L, Peng J, Wong K K Y. Silver nanoparticle-coated suture effectively reduces inflammation and improves mechanical strength at intestinal anastomosis in mice. *J Pediatr Surg* **49**(4): 606–613 (2014)

[12] Li Y, Kumar K N, Dabkowski J M, Corrigan M, Scott R W, Nusslein K, Tew G N. New bactericidal surgical suture coating. *Langmuir* **28**(33): 12134–12139 (2012)

[13] Chen X J, Hou D D, Wang L, Zhang Q, Zou J H, Sun G. Antibacterial surgical silk sutures using a high-performance slow-release carrier coating system. *ACS Appl Mater Interfaces* **7**(40): 22394–22403 (2015)

[14] Walling H W, Christensen D R, Arpey C J, Whitaker D C. Surgical pearl: lubrication of polyglactin suture with antibiotic ointment. *J Am Acad Dermatol* **52**(1): 136–137 (2005)

[15] Chen X J, Hou D D, Tang X Q, Wang L. Quantitative physical and handling characteristics of novel antibacterial braided silk suture materials. *J Mech Behav Biomed Mater* **50**: 160–170 (2015)

[16] Ku S H, Ryu J, Hong S K, Lee H, Park C B. General functionalization route for cell adhesion on non-wetting surfaces. *Biomaterials* **31**(9): 2535–2541 (2010)

[17] Barras A, Lyskawa J, Szunerits S, Woisel P, Boukherroub R. Direct functionalization of nanodiamond particles using dopamine derivatives. *Langmuir* **27**(20): 12451–12457 (2011)

[18] Lee H, Dellatore S M, Miller W M, Messersmith P B. Mussel-inspired surface chemistry for multifunctional coatings. *Science* **318**(5849): 426–430 (2007)

[19] Yan J, Yang L P, Lin M F, Ma J, Lu X H, Lee P S. Polydopamine spheres as active templates for convenient synthesis of various nanostructures. *Small* **9**(4): 596–603 (2013)

[20] Jayakumar R, Prabaharan M, Nair S V, Tokura S, Tamura H, Selvamurugan N. Novel carboxymethyl derivatives of chitin and chitosan materials and their biomedical applications. *Prog Mater Sci* **55**(7): 675–709 (2010)
[21] Viju S, Thilagavathi G. Effect of chitosan coating on the characteristics of silk-braided sutures. *J Ind Text* **42**(3): 256–268 (2013)
[22] Zhang G Q, Ren T H, Lette W, Zeng X Q, van der Heide E. Development of a penetration friction apparatus (PFA) to measure the frictional performance of surgical suture. *J Mech Behav Biomed Mater* **74**: 392–399 (2017)
[23] Shergold O A, Fleck N A. Experimental investigation into the deep penetration of soft solids by sharp and blunt punches, with application to the piercing of skin. *J Biomech Eng* **127**(5): 838–848 (2005)
[24] Azar T, Hayward V. Estimation of the fracture toughness of soft tissue from needle insertion. In *Proceedings of the 4th International Symposium on Biomedical Simulation*, London, UK, 2008: 166–175.
[25] Shergold O A, Fleck N A, Radford D. The uniaxial stress versus strain response of pig skin and silicone rubber at low and high strain rates. *Int J Impact Eng* **32**(9): 1384–1402 (2006)
[26] Peng F, Olson J R, Shaw M T, Wei M. Influence of pretreatment on the surface characteristics of PLLA fibers and subsequent hydroxyapatite coating. *J Biomed Mater Res B Appl Biomater* **88B**(1): 220–229 (2009)
[27] De Simone S, Gallo A L, Paladini F, Sannino A, Pollini M. Development of silver nano-coatings on silk sutures as a novel approach against surgical infections. *J Mater Sci: Mater Med* **25**(9): 2205–2214 (2014)
[28] Cohen R E, Hooley C J, McCrum N G. Viscoelastic creep of collagenous tissue. *J Biomech* **9**(4): 175–184 (1976)
[29] Chen B S, Gu K C, Fang J H, Wu J, Wang J, Zhang N. Tribological characteristics of monodispersed cerium borate nanospheres in biodegradable rapeseed oil lubricant. *Appl Surf Sci* **353**: 326–332 (2015)
[30] Wei Q, Zhang F L, Li J, Li B J, Zhao C S. Oxidant-induced dopamine polymerization for multifunctional coatings. *Polym Chem* **1**(9): 1430–1433 (2010)
[31] Brugnerotto J, Lizardi J, Goycoolea F M, Argüelles-Monal W, Desbrières J, Rinaudo M. An infrared investigation in relation with chitin and chitosan characterization. *Polymer* **42**(8): 3569–3580 (2001)
[32] Chen X G, Park H J. Chemical characteristics of *O*-carboxymethyl chitosans related to the preparation conditions. *Carbohydr Polym* **53**(4): 355–359 (2003)
[33] Zheng M L, Han B Q, Yang Y, Liu W S. Synthesis, characterization and biological safety of O-carboxymethyl chitosan used to treat Sarcoma 180 tumor. *Carbohydr Polym* **86**(1): 231–238 (2011)
[34] Rodeheaver G T, Thacker J G, Owen J, Strauss M, Masterson T, Edlich R F. Knotting and handling characteristics of coated synthetic absorbable sutures. *J Surg Res* **35**(6): 525–530 (1983)

Xiangqiong ZENG. She received her master degree in applied chemistry in 2003 and Ph.D degree in material science in 2006 from Shanghai Jiao Tong University. Then, she worked at Johnson & Johnson consumer groupduring 2006−2010 as staff scientist on skin care technology. From 2011−2015, she was appointed as a tenure track assistant professor at the University of Twente, working on skin tribology. Currently, she is a full professor at Shanghai Advanced Research Institute, Chinese Academy of Sciences. Her research interests are functional interface materials for biotribology and aqueous lubrication, including active control of friction and wear by surface and interface design of skin contacting materials and medical devices, by test methodology development with the design of instrument and bio-inspired human tissue model, and by additive and emulsion development for aqueous lubrication.

Gangqing ZHANG. He received his master degree in chemistry and chemical Engineering in 2013 from Qingdao University, Qingdao, China. After then, he was a Ph.D student in Key Laboratory of Thin Film and Microfabrication Technology (Ministry of Education) at Shanghai Jiao Tong University, China. Meanwhile, he was also a Ph.D student in mechanical engineering at University of Twente, Enschede, the Netherlands. He has recently obtained his Ph.D degree in the two universities. His research interests include biotribology of medical material and development of lubricants, and development of medical biomaterials.