Frictional characteristics of clamp surfaces of aneurysm clips finished by laser processing

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Abstract
An aneurysm clip is a medical instrument that is used intraoperatively to clip a ruptured cerebral aneurysm and reduce the risk of rebleeding. To prevent the clips from slipping off the aneurysm neck, it is very important to maintain a constant clamp force. A high frictional coefficient of the clip blades will also help prevent slippage between the clip blades and the blood vessel. In this study, to raise the frictional coefficients of the clip blades, we used a laser processing machine to produce clamp surfaces of the aneurysm clips with micro-dimples or micro-grooves. The static and dynamic frictional characteristics of the clamp surfaces made in this way were examined. The grooved surfaces with a width of 30 μm and a groove pitch of 40 μm showed the highest frictional coefficient. However, the dimpled surfaces with a shallow depth of 1 μm showed lower frictional coefficients than existing aneurysm clips.

Key words : Aneurysm clip, Clamp surfaces, Frictional coefficient, Laser texturing, Dimples, Grooves, Micro-slip

1. Introduction

A cerebral aneurysm is the dilation, bulging, or ballooning out of part of the wall of an artery in the brain. Rupture of a cerebral aneurysm represents a serious life-threatening condition. An aneurysm clip is a medical instrument that is used intraoperatively to clip a ruptured cerebral aneurysm and reduce the risk of rebleeding, as shown in Fig. 1(a).

Figures 1(b) and 1(c) show existing cerebral aneurysm clips, made of Ti alloy and Co-Cr alloy, respectively. The Co-Cr alloy has superior spring characteristics to the Ti alloy. However, artifacts caused by the Co-Cr alloy tend to be larger in size and effect than those of Ti alloy in postoperative CT or MRI assessment. Therefore, Ti alloy clips are used more widely than Co-Cr alloy clips because of the reduced artifacts. To prevent the clips from slipping off the blood vessels, a number of pyramidal depressions are formed on their clamp surfaces that come into contact with the arteries, as shown in Figs. 1(d) and 1(e). It is very difficult to redesign the shapes and sizes of the pyramidal depressions on the clamp surfaces because they are formed by a metal stamping process. It is very costly to produce various types of hard metal die for the metal stamping process. From a tribological viewpoint, it has not been verified whether the existing design of pyramidal depressions is the optimal form to raise the frictional coefficient of aneurysm clips.

Postoperative angiography showed that the clip can slip off the aneurysm neck (Drake and Allcock, 1973). Some late follow-ups of cerebral aneurysms after neck clipping revealed that even though postoperative angiograms confirmed aneurysm obliteration, clips can slip or break, and the aneurysms can regrow, with hemorrhage recurring several months to years after treatment (de Sousa, 2010), (El-Beltagy, et al., 2010). In addition to slippage, the cerebral aneurysm clip “scissoring” phenomenon is known to occur due to twisting of the aneurysm clip blades during surgery (Hirashima, et al., 2002), (Horiuchi, et al., 2012). Blade scissoring involves slippage of the blades against the artery, and can be prevented by a high friction coefficient of the blades.

To prevent the clips from slipping off the aneurysm neck, it is important to keep the clamp force of the clips...
Table 1 Sizes of dimples and grooves on the clamp surface processed by the laser machine

<table>
<thead>
<tr>
<th>Material</th>
<th>Texture pattern</th>
<th>Diameter or Width (µm)</th>
<th>Pitch (µm)</th>
<th>Depth (µm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ti-6Al-4V alloy</td>
<td>Dimples</td>
<td>30, 40, 50, 60, 100</td>
<td>38</td>
<td>100</td>
</tr>
<tr>
<td></td>
<td></td>
<td>38</td>
<td>100</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>54</td>
<td>100</td>
<td></td>
</tr>
<tr>
<td>Co-Cr alloy</td>
<td>Dimples</td>
<td>30</td>
<td>40</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Grooves (90°)</td>
<td>32, 60, 100</td>
<td>47</td>
<td>100</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>48</td>
<td>100</td>
</tr>
<tr>
<td></td>
<td>Grooves (45°)</td>
<td>32, 60, 100</td>
<td>47</td>
<td>100</td>
</tr>
</tbody>
</table>

Constant (Nagatani, et al., 1998), (Takikawa, et al., 1997), (Woods, 2000). High frictional coefficients of the clip blades will help prevent slippage between the blades and the blood vessels. Therefore, a study to raise frictional coefficients between the blades and the blood vessels is important to ensure the safety of aneurysm clips. Obviously, static frictional coefficients play a major role in preventing slippage of the clips, not only at the time of clamping in surgery but also over the long term. When the blood vessel is clamped with the aneurysm clip during surgery, relative motion will occur between them. Thus, not only static friction but also dynamic friction should be taken into consideration. In the field of tribology, surface texturing is one friction reduction technique applicable in mixed lubrication regimes. There have

![Fig. 1 Cerebral aneurysm and aneurysm clips. (a) Aneurysm clip used to prevent cerebral aneurysm rupture. (b) Aneurysm clips made of Ti alloy with fewer artifacts. (c) Aneurysm clips made of Co-Cr alloy with superior spring characteristics to other biocompatible materials. (d) pyramidal depressions formed on the clamp surface of the aneurysm clip by metal stamping process. (e) Shapes and sizes of depressions.](image-url)
been many studies regarding surface texturing (Etsion, 2005), (Etsion and Sher, 2009), (Nanbu, et al., 2008), (Shinkarenko, et al., 2009), (Wang, et al., 2003). However, there has been comparatively little research relating to an increase in friction by surface texturing or surface roughness, (Wen-Ruey Chang, 1998 and 2002).

In a previous study (Nitta, et al., 2010), clamping forces of the aneurysm clips were analyzed from the view point of strength of materials and the frictional coefficients of the clip blades finished by laser processing were measured. The laser processed blades showed higher frictional coefficient than the existing clip blades. The surface texturing of clip blades is a valuable technique to raise the frictional coefficients between sliding surfaces, resulting in prevention of blade slippage as well as scissoring. Unlike the metal stamping process, with laser processing it is easy to form a variety of shapes of micro-dimples or micro-grooves on the clamp surface in a short time.

In this paper, the effects of surface texturing on the frictional coefficients of the blade surfaces were examined under various dimple depths. Depending on the dimple depth, the frictional coefficients didn’t always increase. With a shallow dimple, the frictional coefficient became lower than that of clip blade with no surface texturing. In addition, the effects of sliding speed on the frictional coefficients were also considered from the view point of visco-elastic properties of silicone tubing which is a counter surface against the clip blade. The variation of the frictional coefficients with the sliding speed was almost similar to that of dynamic Young’s modulus of the silicone tubing.

2. Experimental details

![Fig. 2 Shapes and sizes of test specimens. (a) Ti alloy or Co-Cr alloy. (b) Silicone tubing.](image-url)

![Fig. 3 Clamp surfaces of titanium alloy processed by the laser processing machine. (a) Burrs after laser irradiation and polishing with #1500 emery paper to remove burrs. (b) Dimples. (c) Grooves (90°). (d) Grooves (45°).](image-url)
The clip specimens were made of Co-Cr alloy or Ti alloy. The Co-Cr alloy has superior spring characteristics to other biocompatible materials. However, artifacts caused by Co-Cr alloy tend to be larger in size and effect than those of Ti alloy in postoperative CT and MRI assessments. Therefore, Ti alloy clips are more widely used than Co-Cr alloy clips because of the reduced artifacts.

The shape and size of a test specimen are shown in Fig. 2, which mimics only the blade part of the aneurysm clip, as shown in Fig. 1(b) or 1(c). The sliding region of the test specimen was 1 mm in width and 10 mm in length. Before laser processing, the specimen surfaces were polished to mirror surfaces. The laser processing machine used consists of a Q-switched laser beam generator with a wavelength of 355 nm, two galvano mirrors to rapidly scan the laser beam, and an f0 lens to focus the laser beam on the surfaces of the test specimens. The average power of the third harmonic beam output from the wavelength conversion device is approximately 3 W.

Figure 3 shows a 3D surface profile of a micro-dimple just after laser beam irradiation. Burrs can be seen around the periphery of the micro-dimple. The height of the burrs ranged from 5 μm to 10 μm. These burrs will penetrate into the cerebral artery and may cause rupture. Therefore, after laser irradiation the specimen surfaces were polished with emery paper of #1500 or #4000 to remove burrs. Before the friction tests, the specimens were cleaned in acetone by ultrasonication for 1 minute.

All micro-dimples and micro-grooves were measured with a stylus profilometer. Figures 3(b) – (d) show typical micro-dimples and micro-grooves on the clip specimens under an optical microscope. The diameters of the micro-dimples ranged from 30 μm to 54 μm, the micro-grooves varied in width from 30 μm to 50 μm, and both depressions were about 13 μm in depth. These micro-dimples or micro-grooves were made on the specimen surfaces with a variety of intervals from 40 μm to 100 μm. Table 1 summarizes the sizes of the micro-dimples and micro-grooves processed in this study for both Ti and Co-Cr alloys.

The blade parts were cut out from the existing clips made of Co-Cr alloy and used in the friction test to compare the friction coefficients of the laser-processed specimens. The sizes of the cut blade parts were almost the same as the test specimens, as shown in Fig. 2(a). Pyramidal depressions, 60 μm in depth and 230 μm in length of the side of the square, were observed on the surface of the cut blade.

Counterspecimens against the clip specimens were translucent silicone tubing rather than human blood vessels. The silicone tubing was 3 mm in outer diameter and 0.5 mm in thickness. Young’s moduli of the arteries range in value from 1 to 9 MPa in the literature (Gao, et al., 2006), (Horiuchi, et al., 2012), (Hučko, 2010), (Khamdaeng, et al., 2012). The static Young’s modulus of the silicone tubing used was about 4 MPa and the hardness was 56 Shore A. Figure 4 shows a schematic diagram of the test apparatus for the frictional coefficients. The wall of the silicone tubing was cut in the longitudinal direction and spread open. The spread silicone tubing was fixed on the moving stage with its ends clamped and the outer wall upward. When a pig artery was used instead of the silicone tubing, it was fixed in the same way. The clip specimen was set over the silicone tubing in an acrylic tray, with the longitudinal direction of the clip specimen at a right angle to that of the silicone tubing. In the friction test, the clip specimen was stationary and the silicone tubing was moved in the longitudinal direction at a constant speed under a given normal load. To check for reproducibility, the friction tests were repeated five times under the same conditions and the averaged values were used as the test results.
To ensure the surgical field of view, the aneurysm to be clipped was rinsed with warm saline when surgical clipping was performed. There was almost no body fluid on the aneurysm before surgical clipping. Thus, the friction tests were carried out in physiological saline solution maintained at a constant temperature of 37°C with an electric heater and thermocouple. The sliding speed of the clip specimens was kept constant at 10 mm/s during the experiments, except for those experiments in which the effects of sliding speed on the frictional coefficients were examined. The sliding distance was 5 mm. A dead weight of 1.5 N, which is the same clamping force exerted on the aneurysm clips, was applied to the clip specimens.

3. Experimental results

Figure 5 shows typical variation of frictional coefficient as a function of sliding distance. At around 0.6 mm, the frictional coefficient reached a maximum value. We defined this maximum value as a static frictional coefficient. Subsequently, the frictional coefficient oscillated with time and the oscillation reached a steady state. We defined the average value of frictional coefficients corresponding to this portion as the dynamic frictional coefficient. The frictional force needed to initiate sliding is usually greater than that necessary to maintain it, and hence the static frictional coefficient is greater than the dynamic frictional coefficient.

First, the effect of dimple size on the frictional coefficient was examined. Three different sizes of micro-dimples aligned in a regular lattice were processed on the Ti alloy specimen by defocused laser beams. Figure 6 shows the static and dynamic frictional coefficients as a function of diameter of the micro-dimple with a dimple pitch of 100 μm. Both static and dynamic frictional coefficients increased slightly with increasing micro-dimple diameter although the dimple sizes were changed by only twofold because of limitations of laser defocusing. In the case of micro-grooves, the same tendency of a small effect of groove width on the frictional coefficient was observed. In the case of Co-Cr alloy, the experimental results were almost the same as those for the Ti alloy.
Figure 7 shows the effects of the pitch of micro-dimples and micro-grooves on the frictional coefficients. Both the diameters of the micro-dimples and the groove widths were kept constant at 30 μm. The pitches of the micro-dimples and the micro-grooves were varied from 40 μm to 100 μm. The frictional coefficients decreased linearly with increasing pitch. The static frictional coefficients of the micro-grooves were greater by almost 50% than those of the micro-dimples. These experimental results indicated that the micro-grooves are suitable for surface texturing of aneurysm clips to prevent them from slipping. The dynamic frictional coefficients showed the same tendency as the static frictional coefficients.

Figure 8 shows a comparison of frictional coefficients between groove directions of 90° and 45°, which represent angles of grooves to the sliding direction; the grooves with an angle of 90° are perpendicular to the sliding direction, as shown in Fig. 3(c). Both behaviors of the frictional coefficients were similar to each other and both textures showed higher values of frictional coefficients, although the dynamic frictional coefficients for 90° were slightly larger than those for 45°. The groove direction had little effect on the frictional coefficients.

The clip surfaces grooved with a pitch of 40 μm and groove direction of 90° showed the highest static frictional coefficient. Figure 9 shows such high frictional coefficients together with the frictional behaviors of both micro-dimples with pitch of 40 μm and an existing aneurysm clip. The static frictional coefficient of the aneurysm clip with pyramidal depressions currently in use is about 0.3 and the dynamic frictional coefficient is about 0.25. The grooved surface showed an increase in static frictional coefficient to about 0.75 and the dynamic frictional coefficient was increased to about 0.5. It can be clearly seen that there is room for improvement with respect to the frictional characteristics of the aneurysm clips that are in current use.

Figures 10(a) and 10(b) show the effects of sliding speed on the static and dynamic frictional coefficients. The
sliding speed was varied from 0.5 mm/s to 15 mm/s. The frictional coefficients increased rapidly with increasing sliding speed and then leveled off except for the static frictional coefficient of the grooved surfaces. The sliding speed had a large effect on the frictional coefficients. However, the superiority of grooves did not change regardless of sliding speed.

To examine whether the frictional behaviors of the silicone tubing are similar to those of blood vessels, a pig artery about 4 mm in diameter and 0.2 mm in wall thickness was used for the friction test. Figure 11 shows a comparison of the static frictional coefficients of the blood vessel and silicone tubing. The pig artery was cut and spread open in the same way as the silicone tubing. In the case of the grooved surfaces, the static friction was similar for both pitches of 40 μm and 100 μm. On the other hand, in the case of the dimpled surfaces, the reduction rate of the frictional coefficient with pitch was greater with the pig artery than the silicone tubing.

Figure 12 shows a comparison of the frictional coefficients between the silicone tubing and pig artery when the sliding speed was changed. The dashed lines represent the experimental data of the silicone tubing. In the cases of the
existing clips and the grooved surfaces, the frictional coefficients against the silicon tubing were similar to those against the blood vessel. In the case of the dimpled surfaces, the frictional coefficients against the silicon tubing were smaller than those against the blood vessel. However, the frictional behaviors as a function of sliding speed were similar to each other. These observations indicated that the silicone tubing is applicable to check the frictional characteristics of blood vessels.

The frictional coefficients increased markedly with increasing sliding speed. We postulated that one reason for this was the viscoelastic characteristic of the silicone tubing used in this study. The viscoelastic characteristics of the storage and loss moduli were measured with a rheometer. Cyclic tension was applied to the silicone tubing with a frequency sweep up to 100 Hz. From these results, the dynamic Young’s modulus, $E(t)$, as a function of time was obtained as described in our previous study (Nitta and Terao, 2003).

$$
E(t) = E'(\omega) - 0.4E''(0.4\omega) + 0.014E''(10\omega)
$$

Where $t$ is time [s] and $\omega$ is angular frequency [rad/s]. The range of $\omega$ was from 0.1 Hz to 100 Hz. According to Eq. (1), the dynamic Young’s modulus at a given time can be calculated. The time at a given sliding speed is determined by dividing the representative length of the dimples or grooves, in this case roughly 10 μm, by the sliding speed.

Figure 13 shows the dynamic Young’s modulus of the silicone tubing together with the dynamic frictional coefficients of the dimpled and grooved surfaces. As shown in the figure, the variations in dynamic Young’s modulus with the sliding speed were similar to those of the frictional coefficients. Thus, there is a strong correlation between the viscoelastic characteristics of the silicone tubing and the frictional coefficients.
Surface texturing is one friction reduction technique in mixed lubrication regimes. However, the surface texturing technique raised the frictional coefficients of the aneurysm clip surfaces in this study. Shinkarenko et al. reported that there is an optimum aspect ratio, i.e., the ratio of dimple depth to dimple diameter, to minimize the frictional coefficients (Shinkarenko, et al., 2009). In their study, the optimum aspect ratios ranged from 0.05 to 0.1. We used dimples with aspect ratios from 0.21 to 0.43, which are larger than the optimum values. Therefore, we prepared various dimples of ranging in depth from 0.5 μm to 9 μm. Figure 14 shows the measured frictional coefficients as a function of the dimple depth, together with SEM images of dimples 1 μm and 9 μm in depth. In this figure, a depth of 0 μm means that no laser texturing was applied to the specimen surface. The minimum frictional coefficients appeared at a dimple depth of 1.0 μm, corresponding to a dimple aspect ratio of 0.05, which is coincident with the optimum dimple ratio in the study of Shinkarenko et al. (2009). This is because the dimple surface of the optimum aspect results in more efficient hydrodynamic pressure generation in the mixed lubrication regime. In the case of a dimple depth of 0.5 μm, both the static and dynamic frictional coefficients were almost same as those of non-textured surfaces. These observations suggested that such shallow dimples could not generate efficient hydrodynamic pressure.

Next, we examined the effects of dimple pitch on the frictional coefficients on the clamp surfaces. The dimple depth formed was 1 μm or 9 μm. Figure 15 shows the frictional coefficients as a function of the dimple pitch from 30 μm to 100 μm. In the case of a dimple depth of 9 μm, the frictional coefficients decreased linearly with dimple pitch.

Fig. 14 Variation of static and dynamic frictional coefficients with dimple depth for Ti alloy (dimple diameter and pitch: 20 μm and 40 μm, respectively).

Fig. 15 Variation of static and dynamic frictional coefficients with dimple pitch for Ti alloy (dimple depth: 1 μm or 9 μm).
and leveled off over a dimple depth of 70 μm. Below the dimple depth of 70 μm, there was a somewhat larger variation of the static frictional coefficient, and such larger variation almost disappeared over a dimple depth of 70 μm. This variation in static frictional coefficient had a strong association with unstable and irregular solid contact at the peripheries of the dimples. On the other hand, the frictional coefficients of a dimple depth of 1 μm decreased initially and then increased linearly with dimple pitch. They leveled off over a dimple depth of 70 μm. The variation of the static frictional coefficient became smaller than that with a dimple depth of 9 μm.

The highest frictional coefficient was obtained with a dimple depth of 9 μm and dimple pitch of 30 μm. However, if the dimple depth was as shallow as 1 μm, the frictional coefficient became lower, resulting in a dangerous situation for the aneurysm clips.

4. Discussion

To prevent aneurysm clips from slipping off the blood vessels, a number of pyramidal depressions are formed on the clamp surfaces of existing aneurysm clips. However, postoperative angiography showed that an aneurysm clip can slip off the aneurysm neck, leading to an increased risk of rebleeding. Therefore, it is important to raise the frictional coefficient of the clamp surfaces of aneurysm clips to enhance their safety.

To increase the frictional forces of the clamp surfaces, the normal force, such as clamping force, and/or the frictional coefficient should be increased. However, there are limits to increasing the clamping force because excess clamping force will cause failure of the aneurysms. Indeed, the clamping force at a given position is kept at a constant optimum value in the manufacturing process. Therefore, we were unable to increase the clamping force. Therefore, increasing the frictional coefficient is the only effective method of raising the frictional force of aneurysm clips.

In this study, we used the laser-processing technique to prepare the clamp surfaces of aneurysm clips with micro-dimples or micro-grooves and the static and dynamic frictional characteristics of such clamp surfaces were examined. The grooved surfaces with a groove width of 30 μm and groove pitch of 40 μm showed the highest frictional coefficient, which was greater than that of existing aneurysm clips with pyramidal depressions. Thus, we were able to demonstrate the feasibility of laser surface texturing for significantly improving the tribological performance of aneurysm clips.

In general, the static frictional force between rubber and a hard flat surface has two contributions commonly described as the adhesion and hysteresis components (Persson, 1998). The hysteresis component results from the internal friction of the rubber, or the silicone tubing as used in this study. For the static tangential force, which is quasi-statically applied, the hysteresis component will be negligible. Therefore, only the adhesive component remains, which is estimated by multiplying the real contact area and the shear strength per unit area between the rubber and the hard flat surface. As the shear strength of the adhesive junction can be assumed to be constant, the total amount of real contact area will determine the frictional force of the aneurysm clip. Therefore, further studies involving measurement of real contact area between the silicone tubing and the clip blade with or without laser texturing are required. For example, the real contact area will increase in the outer peripheries of the dimples because of stress concentration, resulting in increases in the static frictional forces. Ploughing the silicone tubing at the edges of the deep grooves or dimples may also occur.

Generally, precision measurement of real contact area over the whole apparent contact area of actual machine elements is difficult, because of the narrow field of view of optical microscopes, and few such measurements have been reported. In a previous study, we developed a novel measurement method for real contact area over a large apparent contact area (Nitta, et al., 2013), where the total amount of real contact area could be measured effectively over the
whole apparent contact area between a rubber pick-up roller with grooves and a glass plate, and the results showed that the static frictional force was proportional to the total amount of real contact area between them.

Dimpled surfaces with a shallow depth of 1 μm showed lower frictional coefficients than existing aneurysm clips. In tribological studies of the mixed lubrication regime, the surface texturing technique is one friction reduction technique and has been used in many previous studies. Shinkarenko et al. reported out that there is an optimum ratio of dimple depth to dimple diameter to minimize the frictional coefficients (Shinkarenko, et al., 2009). In this study, shallow dimples with a depth of 1 μm lay in the range of this optimum ratio. Therefore, special care should be taken regarding the ratio of dimple depth to dimple diameter in the laser texturing process for the safety of aneurysm clips.

In the laser texturing process, micro-dimples or micro-grooves were formed on the specimen surfaces just after laser beam irradiation. As shown in Fig. 3(a), burrs also occurred around the periphery of the micro-dimples. In this study, the height of the burrs ranged from 5 μm to 10 μm. We felt that these burrs would penetrate into the cerebral arteries and may cause them to rupture. Therefore, such burrs were removed by polishing specimen surfaces with emery paper after laser irradiation. However, the risk of artery rupture by penetration of such short burrs would be quite small. Thus, the frictional coefficients between clamp surfaces with burrs and silicone tubing could be increased in ploughing mode in which the surfaces of the silicone tubing are pushed along ahead of the burrs. In future, it will be very important to study the safe burr height that will not harm the arteries but raise the frictional coefficients.

5. Conclusions

In this study, we used a laser processing machine to produce aneurysm clip clamp surfaces with micro-dimples or micro-grooves. The static and dynamic frictional characteristics of the clamp surfaces made in this way were examined. The experimental results demonstrated that circular depressions or small grooves on the clamp surfaces were applicable to raise the frictional coefficients and prevent the clips from slipping off the blood vessels. The grooved surfaces with a width of 30 μm and a groove pitch of 40 μm showed the highest frictional coefficient. However, the dimpled surfaces with a shallow depth of 1 μm showed lower frictional coefficients than existing aneurysm clips.

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