High Tear Strength Artificial Suture Pads for Hand Skin Model from Degummed Silk Fiber-Reinforced Polydimethylsiloxane

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Abstract

The objective of this research is to develop a high tear strength suture pads from a polydimethylsiloxane (PDMS) composite reinforced with degummed silk fibers (DSFs) as artificial hand skin for suturing practice. The influences of fiber diameters, lengths and contents of DSFs on mechanical properties of the prepared PDMS/DSFs composites were systematically investigated. In composite preparation, PDMS resin and silicone oil were mixed, followed by the gradual addition of DSFs with continuous stirring. The uncured mixture was then cast in molds and cured at room temperature for 4 hours. The results demonstrated that DSFs with a smaller diameter of 170 μm significantly improved tensile strength, elastic modulus, hardness, and, most notably, tear strength—an important property for resisting cuts from medical sutures. The morphology of the composites revealed that the DSFs have uniform dispersion and good adhesion in the PDMS matrix. Additionally, the lengths and contents of DFSs contributed to the enhancement of the mechanical properties of the PDMS/DSFs composites. The developed PDMS/DSFs composites exhibited mechanical properties and texture relatively close to those of real human skin as evaluated by experienced orthopedic surgeons. This indicates that a PDMS/DSFs composite is an effective artificial hand skin model for suture practice.

Keywords: Polymer composites, Mechanical properties, Natural fiber, PDMS/DSFs composites, Artificial hand skin

1. Introduction

Medical simulators are an educational tool that is highly effective in training medical personnel, in order to develop the necessary skills and techniques before working with a patient. The medical simulator is used in all areas of medical education including birthing simulator for delivery of baby,[1] for self-training in the early detection of breast cancer,[2] suture pad for practicing suture wounds,[3] and so forth. However, those surgical training model simulators are not widely available, because of their high cost. These commercialized products give different human liked skin, which then affects practicing.

Currently, the practicing experience for hand surgery of a child with congenital limb abnormalities has high demand. Nevertheless, none of artificial children hand skin model giving human liked skin are available in the market leading to poor practicing experience of surgery. As a result, inexperienced surgeons gain surgical experience only from actual surgeries, which increases the risk of surgical scars or incomplete procedures. Rubber gloves were reported to be used as congenital hand anomalies simulator for train inexperienced surgeons.[4] Therefore, this research aims to fabricate artificial hand skin models for surgical practices. However, the structure of human skin differs in many variables including age, sex, and location.[5,6] Surgery of the child's hand usually uses the web space of the hand as tissue and sutures in which it can be compared to the abdomen and thigh which has high flexibility and high tear strength.[7] Consequently, the simulation of hand skin requires a material that is flexible and similar to human skin, such as polydimethylsiloxane (PDMS). Due to these advantages, a large number of artificial products made from PDMS are commercially available with different purposes for example medical training simulators including orthopedic surgery model, pediatric model, and forensic model under the BM Simulator and female right hand under DreamMoulderStudio.

PDMS is an elastomer containing Si-O bonds in its backbone and two methyl groups in repeating structural unit[8] having special features i.e. high flexibility, high thermal stability, hydrophobicity, and biocompatibility.[9,10,11,12] However, it exhibits poor mechanical properties and tear resistance, which need improvement.[13,14] Arm *et al*., (2019) reported the influence of an amount of polyester fibers as reinforcing agent on mechanical properties of PDMS. Higher fibers content resulted in higher modulus and hardness.[14] Moreover, Prasomsin *et al*. (2021) developed a suture pad for medical training in which mechanical properties were enhanced by increasing the fibers length.[3] This is because a polymer matrix can be strengthened against a tension load more effectively with longer fibers.

Natural fibers are gaining attention as a substitute for synthetic fibers, because they are environmentally friendly and biodegradable.[15,16,17,18,19] Cellulose, hemicelluloses, lignin, pectin, and wax are main constituents of the natural fibers, making them hydrophilic, resulting in poor wettability to PDMS matrix.[20,21,22,23,24] In this research, degummed silk fibers (DSFs) were used to improve the mechanical properties of PDMS. Silk fibers are natural fibers which are spun as a natural process of worms. About 25% of the raw silk consists of undesirable constituents which is known as "silk gum" or "sericin". Sericin is a sticky substance produced by the silkworm that holds the strands of silk together. The process of removal of this gum from silk fibers is called "degumming". Degumming of silk improves the sheen, color, hand and texture of the silk fibers. These also have excellent mechanical properties in terms of toughness and strength compared to other natural fibers so they are widely used for medical devices.[25,26] The majority of silk's chemical structure is hydrophobic, resulting in good wettability with silicone matrix and good composite properties.[27]

Adding reinforcing fibers affect not only the properties of PDMS but also affect the rheology of PDMS prepolymer. Prepolymer rheological properties are affected by several factors, including the amount of reinforcing fibers, the interaction of fiber-fiber and fiber-matrix, fiber breakage, and fiber migration. Rheological properties can indicate the final form of a product and are used in considering processability.[28] In addition, the rheological properties can also describe the fibers distribution in the matrix. There are significant distinctions between the physical characteristics of fibers and those of a polymer matrix, which can bring about substantial enhancements in physical and rheological properties of uncured prepolymer.[29] In this research, the mechanical properties of PDMS reinforced with DSFs were improved for use as artificial hand skin having high tear resistance. The effects of fibers diameter, fibers length, and content on mechanical properties were investigated. In addition, the effects of morphological properties of PDMS/DSFs composites and rheological behavior of PDMS prepolymer/DSFs dispersions were also studied.

2. Experimental

2.1 Materials

PDMS prepolymer (RTV 2-230 MW) which can be cured at room temperature, curing agent and silicone oil were supplied by Mat Wealth Co., Ltd., Thailand. DSFs were purchased from Chul Thai Silk Co., Ltd., Thailand.

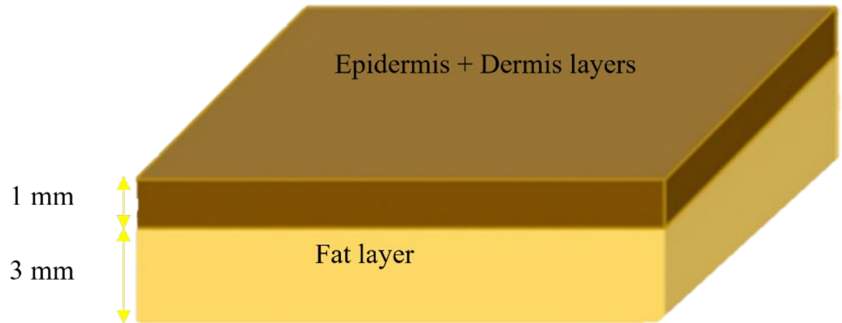
2.2 Preparation of PDMS/DSFs composites

PDMS/DSFs composites were prepared using DSFs with different fibers diameter of 360, 260, 170, or 100 µm, fibers length of 0.5, 1.0, or 1.5 cm, and content of 0.5, 1.0, or 1.5 phr. PDMS resin was mixed with 20 phr of silicone oil and stirred to obtain a homogeneous mixture. 1 phr of curing agent was added into the mixture and immediately stirred. The DSFs were added into the mixture and carefully mixed so that the silk fibers were uniformly dispersed avoiding fracture.

This uncured material was poured into the iron mold and placed into a vacuum oven at a room temperature for 10 min to eliminate bubbles. Then, the samples were cured at a room temperature for 4 hrs. PDMS neat polymer was also prepared by the same process, as the PDMS composite, but without the addition of DSFs.

2.3 Artificial hand skin preparation

The artificial hand skin was prepared into two layers as shown in Scheme 1. The first layer is the skin layer, that mimic epidermis and dermis layers with thickness 1 mm and the second layer is fat layer with thickness 3 mm. The skin layer is a PDMS/DSFs composite that was prepared as previously described in section 2.2. The fat layer was prepared by mixing PDMS with silicone oil at a composition of 1:1 and 1 phr of curing agent. The mixture was room temperature cured for 2 days.



Scheme 1. Artificial hand skin.

2.4 Material characterization

The tensile tests were performed using a Universal Testing Machine (UTM) model 5567 from Instron Co., Ltd., (Bangkok, Thailand). Five samples with dumbbell shape were tested according to ASTM D 412 Type C. A load of 1 kN and a testing speed of 500 mm/min were applied.

The tear tests were performed according to ASTM D 642 Type C. A load of 1 kN and a testing speed of 500 mm/min were applied onto the samples. Five measurements were carried out to obtain average values.

Suture retention strength or suture pullout strength were determined by using a UTM model 5567 from Instron Co., Ltd., (Bangkok, Thailand). The samples were cut into rectangular shape, with dimensions of 1.8 × 2 cm. The surgical suture size 5/0 metric was placed through the sample approximately 2 mm from the edge by the simple loop.[30] A tension test was executed at 1 mm/s until the point of complete suture pullout. The maximum load of suture pullout was recorded for each sample. Five measurements were carried out to obtain average values.

The samples with dimensions of 50 × 50 mm and minimum thickness of 6.4 mm were prepared and tested according to ASTM D 2240 by using Shore A durometer. The average hardness was measured from five-points on the samples.

Morphology, dispersion of DSFs in PDMS matrix and fibers diameter were investigated by scanning electron microscope (SEM) model SU 3500 from Hitachi Co., Ltd., (Tokyo, Japan).

The samples were coated with a thin layer of gold. The average diameter of fibers was measured from five-points diameters from SEM micrographs by using ImageJ software.

The contact angle measurements of water and PDMS on DSFs fibers were conducted according to the procedure reported in Schellbach *et al.* (2016),[31] 170 μm DSFs fiber was selected for the test. Two fibers were placed parallelly and attached on a fiber holder. The bridge between those fibers was kept at 1 mm. Deionized water or PDMS were slowly dropped on two fibers bridge. The image of water or PDMS droplet were taken by microscope. The contact angle measurements were tested three times. The images of droplet were taken as illustrated in Figure 1 and then analyzed using ImageJ software. The contact angle was calculated using the following equation. Diameter (d) and height (h) of the meniscus on at least one side of the droplet column were measured to determine contact angle.

$$\theta = 90 - 2\tan^{-1}(\frac{2h}{d})$$

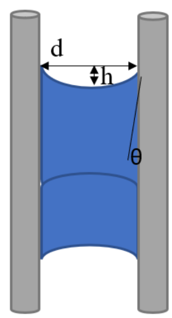


Figure 1. Schematic of a liquid bridge between two fibers of diameter.

Rheological properties of uncured PDMS (prepolymer)/DSFs were performed on a rheometer model HAAKE MARS from Thermo Fisher Scientific Co., Ltd., (Waltham, United States). The samples were prepared on uncured state and tested at steady state shear (ranging from 0.01 to 100 s^{-1}) to study their flow behavior.

3. Results and discussions

3.1 Morphology of DSFs

The morphologies of DSFs were observed by SEM and the obtained micrographs are shown in Figure 2. The fibroin strands were spirally twisted together by the spinning process from the cocoon becoming a silk fiber. This spiral twisting causes the development of mechanical strength by enhancing mechanical interlocking which increased inter-fiber friction and greater fiber entanglements as reported by Zhou et al. (2021).[32] From the micrographs, the diameter of DSFs were measured to be 360±10, 260±10, 170±20, or 100±20 µm, respectively.

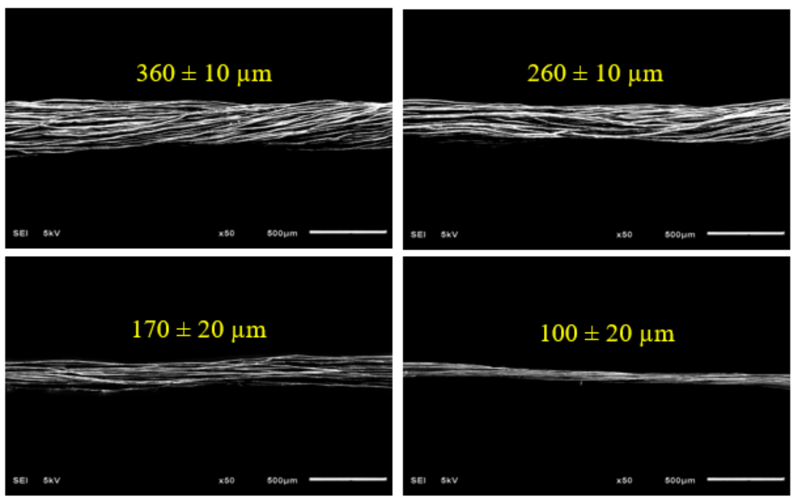


Figure 2. Morphology of degummed silk fibers showing the structure of the silk fibers and the measuring of diameters.

3.2 Effects of DSFs Diameter on Mechanical Properties of PDMS/DSFs composites

3.2.1 Tensile Tests

For the tensile test a tension force is applied to a specimen and the specimen's response to the stress is recorded. From this test the tensile strength, elongation at break, and Young's modulus of specimen are determined. Tensile strength is an important property that should be taken into consideration when mimicking artificial hand skin. The surgical hand model must be as strong as the real human skin, in order to be durable and effective for surgical practice. The tensile strength of PDMS/DSFs with different fibers diameter and a fibers length of 0.5 cm at a constant fibers content of 1 phr, is shown in Figure 3a. The results obtained showed an improvement in the tensile strength of the neat PDMS by incorporation of DSFs. The DSFs contain bulky repetitive hydrophobic domains, interrupted by small hydrophilic groups.[15,33,34] The hydrophobic characteristics of DSFs play an important role in the interaction of fibers with the hydrophobic also PDMS prepolymer (matrix) resulting in good wettability, which means that the liquid resin flow over the fibers covering every dip of their rough surface displacing all air.[35] For this reason, the PDMS/DSFs composites displayed good compatibility and strong adhesion and between DSFs and PDMS matrix. Furthermore, the DSFs consist of β-sheets crystallites, which are the main constituents of their structure, attributing to the strong mechanical properties of the fibers.[36]As a result, the fibers are to withstand more stresses than other natural fibers. On the other hand, Zhang *et al*. (2015) reported the use of highly hydrophilic cellulose fibers for PDMS reinforcement. Their results showed a decrease in tensile strength of the reinforced PDMS which was due to the poor interfacial bonding between the cellulose fibers and PDMS. By this context, the fibers must be chemically treated to improve their surface hydrophobicity.[37] In addition, Chakkour *et al*. (2023) reported most natural fibers are chemically treated before being used as reinforcing fibers to remove their amorphous parts.[38] Therefore, the compatibility of the DSFs and the PDMS matrix improved the mechanical strength of the composites without the need for chemical treatment. Moreover, the tensile strength of PDMS/DSFs composites showed a similar tendency to that reported by Moudood *et al.* (2019);[39] it tends to increase with decreasing fiber diameter. The tensile strength value of the neat PDMS was observed to be 1.27 ± 0.03 MPa, which was then increased to 1.31 ± 0.01, 1.35 ± 0.03, 1.41 ± 0.07, or 1.63 ± 0.04 MPa for the PDMS/DSFs composites reinforced with fibers size of 360, 260, 170, or 100 μm, respectively. The smaller fiber diameter of DFSs enhanced the tensile strength of PDMS/DFS composites because it provided greater yarn strength compared to larger fiber diameters following the equation from Hearle *et al.* (1969).[40]

$$\frac{\text{Yarn strength (or modulus)}}{\text{Fiber strength (or modulus)}} = \frac{E_y}{E_f} = \cos 2\alpha\,(1 - \mathrm{k}\operatorname{cosec}\alpha)$$

Where E_y is yarn strength, E_f is fiber strength, α is the twist angle, and k is express by the following equation:

$$\mathrm{k} = \frac{\sqrt{2}}{3\mathrm{L_f}}\left(\frac{\mathrm{aQ}}{\mu}\right)^{\frac{1}{2}}$$

where L is the fiber length, a is the fiber diameter, Q is the migration period and μ is the coefficient of friction.

From the above equation, the fiber diameter significant effect on K value and the lower the K value the higher the yarn strength; as fiber diameter decreased, the K value decreased leading to higher yarn strength. Additionally, the smaller fibers facilitated better stress transfer between fibers due to the reduced gaps, leading to increased resistance to microcracks at the fiber-polymer matrix interface.[41] The development of a hand surgery model required the improvement of polymer composites to resemble human skin; the tensile strength of the real human abdomen and thighs were reported to have values in the range of 1-2.5 MPa.[42,43] Our results showed that, the tensile strength values of PDMS/DSFs composites were within this range, indicating that these composites are suitable materials to be further developed for use as a hand surgery model for surgical training.

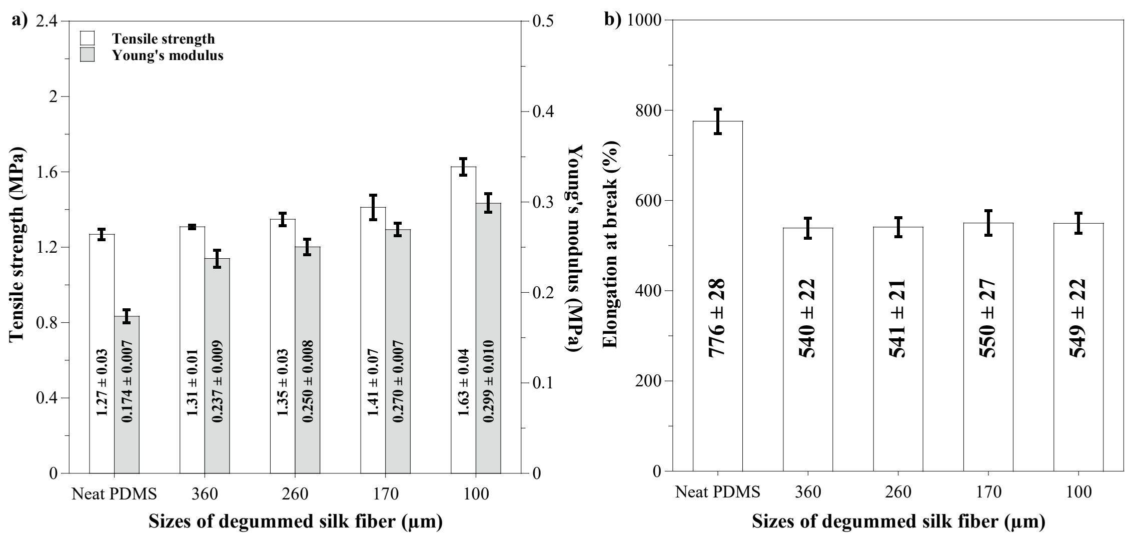


Figure 3. a) Tensile strength and Young's modulus and b) elongation at break of PDMS/DSFs composites with fibers having different diameters and fibers length of 0.5 cm, at a constant fibers content of 1 phr.

The tensile modulus of PDMS/DSFs having different fibers diameter is presented in Figure 3a. The tensile modulus of the neat PDMS was observed to be 0.174 ± 0.007 MPa, which was improved to 0.237 ± 0.009, 0.250 ± 0.008, 0.270 ± 0.007, and 0.299 ± 0.010 MPa for the PDMS/DSFs composites reinforced with fiber sizes of 360, 260, 170, and 100 µm, respectively. This was due to the high stiffness of DSFs embedding in a softer PDMS matrix, resulting in a higher rigidity of the PDMS/DSFs composites.[44] Moreover, the tensile modulus of PDMS/DSFs composites showed a similar trend, as that reported by Abdal-hay et al. (2012),[45] i.e., it tended to increase with decreasing fibers diameter. The maximum modulus of 0.30 MPa was achieved by the PDMS/DSFs composite with 100 μm fiber size. The reasons behind these are same as described earlier for the increment of tensile strength with smaller diameter.

The elongation at break of these composites is illustrated in Figure 3b. The elongation at break of the neat PDMS was about 776 ± 28%. With an incorporation of DSFs the elongation at break of composites decreased in the range of 540-550 ± 21-27%. This is due to the higher semi-crystallinity of silk fibers with elongation at 4-26%.[47] Alam *et al.* (2011) also reported a decrease in elongation at break of PDMS reinforced with DSFs.[47] On the other hand, there are no significant differences in elongation at break of the PDMS/DSFs composites with DSFs with different diameters.

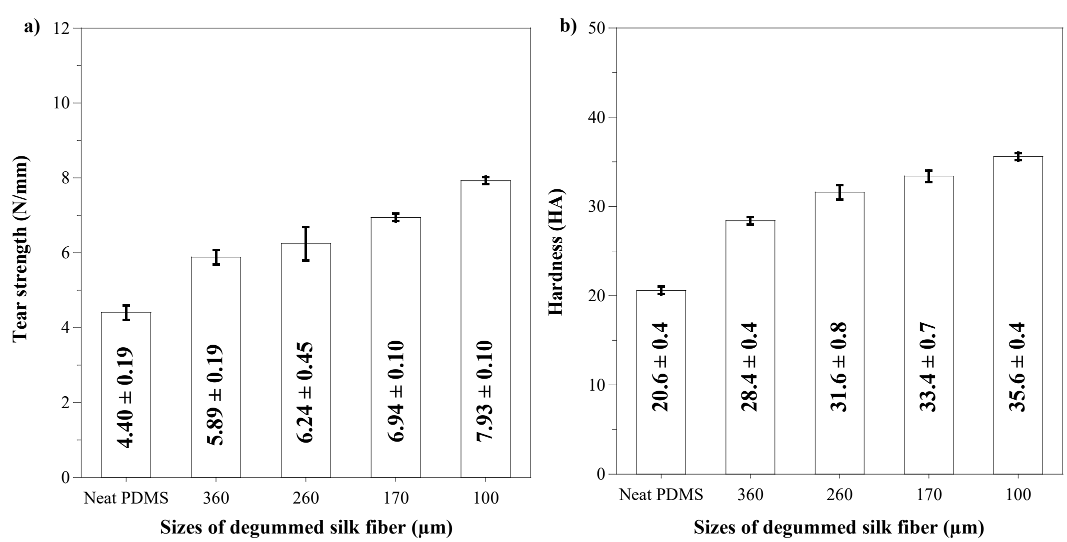


Figure 4. a) Tear strength and b) hardness of PDMS/DSFs with different diameter of fibers and a constant length of 0.5 cm at a constant fibers content of 1 phr.

3.2.2 Tear tests and hardness

The tear strength measures the cut resistance of a material under tension. Tear strength is one crucial variable mechanical property for surgical practice. The surgery involves pulling the skin from one point to cover the wound area; thus, the surface being pulled must have high tear resistance. Tear strength of PDMS/DSFs having DSFs different fibers diameter and a constant fibers length of 0.5 cm, at constant fibers content of 1 phr is shown in Figure 4a. The tear strength value of the neat PDMS was observed to be 4.40 ± 0.19 N/mm. The tear strength of the PDMS/DSFs composites increased significantly compared to the neat PDMS. This is due to the dispersed fibers, which deflect or arrest the growing cracks, resulting in further crack resistance. The tear strength of composites tended to increase with decreasing fibers diameter. The tear strength of the PDMS/DSFs composites were 5.89 ± 0.19, 6.24 ± 0.45, 6.94 ± 0.10, or 7.93 ± 0.10 N/mm for 360, 260, 170 or 100 µm, respectively. The small diameter of DSFs enhanced tear strength due to the tighter packing of smaller fibers, which helps prevent the initiation of interfacial microcracks caused by stress concentration.[41] Additionally, as the fiber diameter decreased, the fiber-matrix interface area increased, leading to more effective stress transfer from the PDMS matrix to the fibers.[41] These factors together contribute to the improvement of tear strength. This explanation is supported by the increased tensile strength observed with smaller fiber diameters. Consequently, it was proposed that the PDMS/DFSs has anisotropic properties where the material exhibits the same characteristics in all directions.

Hardness is one of the key properties used in comparison with real human skin. The surgical hand model must have a hardness value that is close to that of the real human skin to be effective when it is used for surgical practice. The hardness of PDMS/DSFs composites having different fibers diameter and a fibers length of 0.5 cm at a constant fibers content of 1 phr is displayed in Figure 4b. The hardness of the neat PDMS was 20.6 ± 0.4 HA and increased with an addition of DSFs to 28-36 ± 0.4-0.8 HA. This was because of the chemical structure of silk fibroin having H-fibroin chain and L-fibroin chain linked together via disulfide bond at the C-terminus of H-fibroin chain. The H-fibroin chain is formed by a highly repetitive crystallite containing hydrophobic domains; this tends to form **β**-sheets crystallites, through hydrogen bonding and hydrophobic interactions. The **β**-sheet is characterized by the structure of nanocrystals, which plays an important role in achieving the high stiffness of silk fibers.[36] Therefore, reinforcing PDMS with silk fibers, it resulted in a higher hardness of composites. Nevertheless, the hardness is related to elastic modulus i.e., a material with high hardness has also a high modulus.[48] Furthermore, the hardness is another crucial variable used to compare human skin. The average hardness of human skin was reported to be about 30-40 HA.[49] The hardness of PDMS/DSFs composites with characterized as 260, 170, and 100 µm, were within this range and resembled as the real human skin. In conclusion, PDMS/DSFs composites are suitable to mimic real human hand skin for surgical training, in terms of tensile strength and hardness.

3.2.3 Morphology study

The fracture surface of PDMS/DSFs composites was observed by SEM. It can be seen from Figure 5 that the DSFs showed surface roughness. In addition, the DSFs are free of impurities and have hydrophobic characteristics, similar to that of the PDMS matrix; so, the matrix can spread and cover all the surface of DSFs. Hence, the DSFs and the PDMS matrix have mechanical interlocking that leads to strong adhesion of fibers and matrix.[50,51]

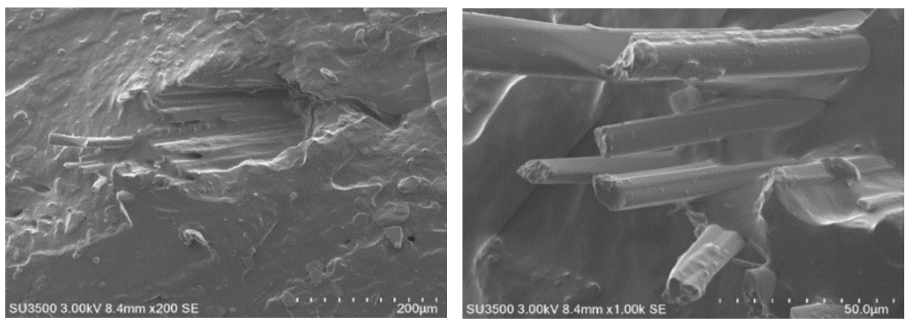


Figure 5. Fracture surface of PDMS/DSFs composites.

The dispersion of DSFs in PDMS is shown in Figure 6. The results showed that, fibers of all diameters within the studied range were evenly distributed throughout the PDMS matrix, indicating that fiber diameter had no impact on their distribution within the composites. Moreover, no break or agglomeration was observed, indicating that a suitable technique for the preparation of PDMS/DSFs composites was used.

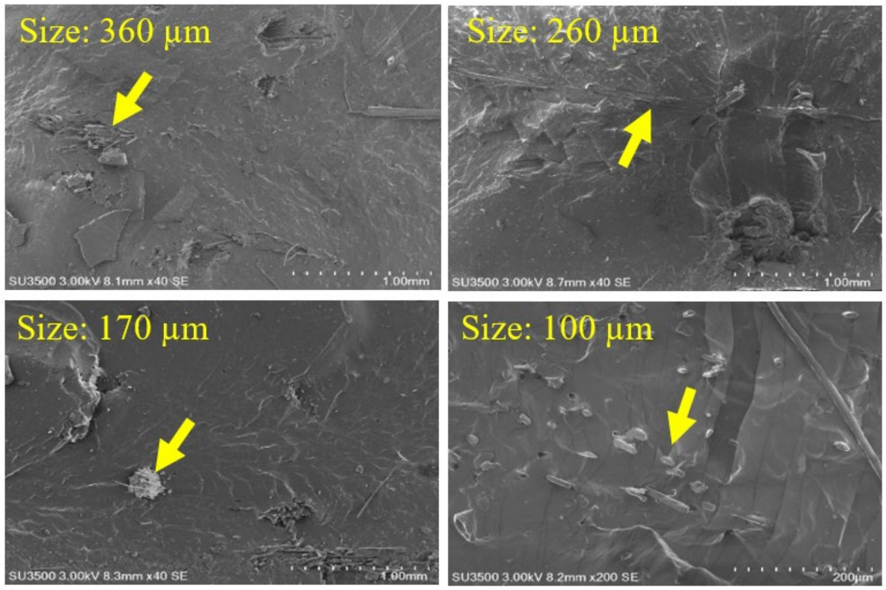


Figure 6. Dispersion of PDMS/DSFs composites contained DSFs with different diameters.

3.3. Effect of DSFs Length on Mechanical Properties of PDMS/DSFs Composites

3.3.1 Tensile tests

The tensile strength of PDMS/DSFs contained DSFs with different fibers length and diameter (360, 260, 170, 100 µm) at constant fiber content of 1 phr, is shown in Figure 7a. The tensile strength of all composites 360, 260, 170, 100 µm increased significantly with increasing the fibers length as also reported by Prasomsin *et al*. (2021) and Wong *et al*. (2010).[3,52] This is due to the fibers which carry the load from the matrix. The force applied to the fibers distributes the stress to the fiber ends. As a result, the longer fibers have a greater load carrying capacity, resulting in increased tensile strength. The maximum tensile strength, in the range of 1.6 to 2.1 MPa was observed for the PDMS/DSFs composite at a fiber length of 1.5 cm. The tensile modulus of all PDMS/DSFs composites (360, 260, 170, 100 µm) exhibited also the same trend as the tensile strength, i.e., its values increase with increasing the fibers length.

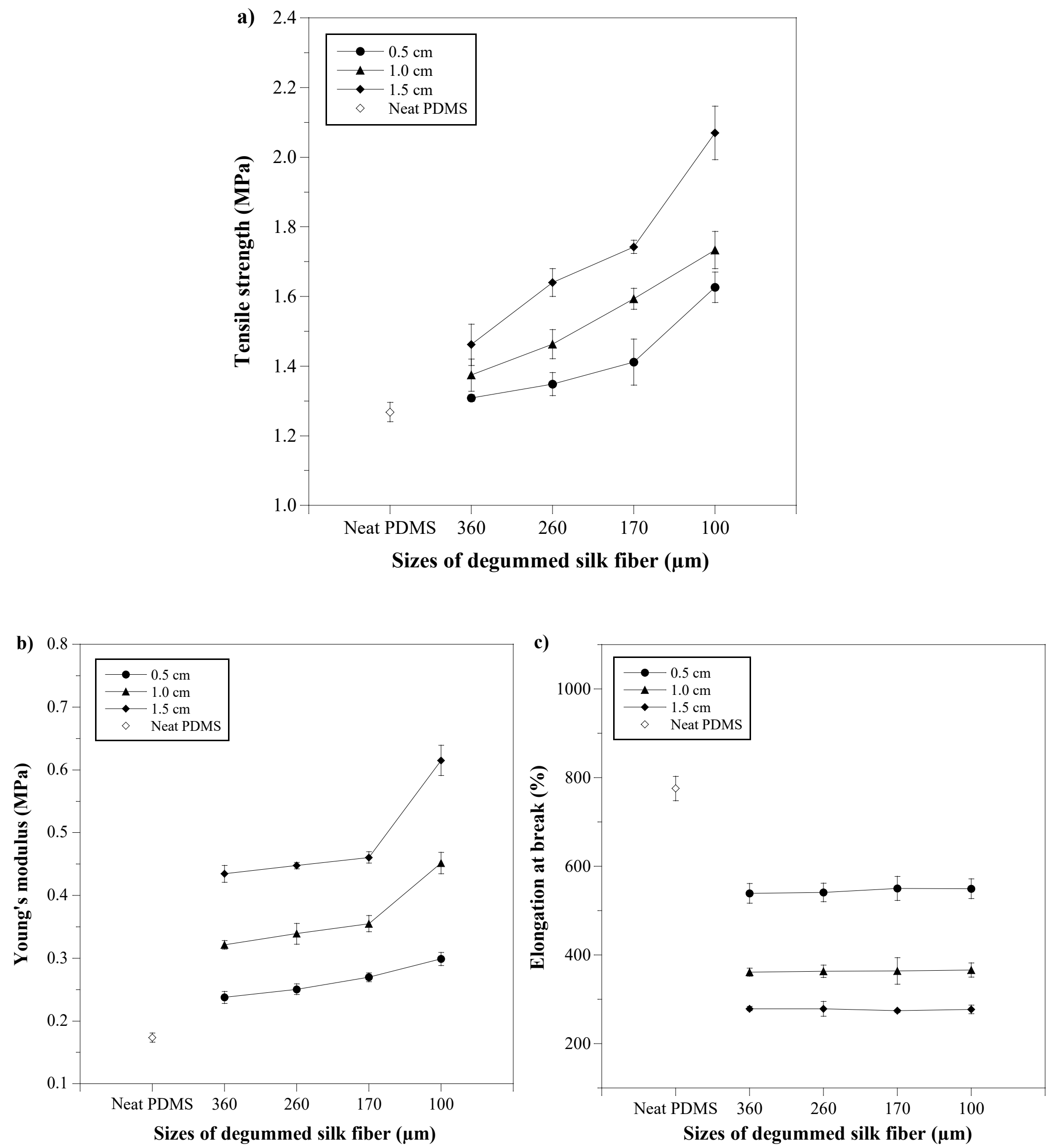


Figure 7. a) Tensile strength, b) Young's modulus, and c) elongation at break of PDMS/DSFs having different fibers length and diameter (360, 260, 170, 100 µm) at constant fibers content of 1 phr.

The elongation at break of PDMS/DSFs composites having different fibers length at constant fiber content of 1 phr is illustrated in Figure 7c**.** The decrease in elongation at break with longer fiber lengths is attributed to the significantly larger surface area of the fibers. For example, with a fixed fiber diameter of 360 µm, the surface areas of fibers with lengths of 0.5, 1.0, and 1.5 cm were calculated (based on a cylindrical shape) to be 12.1, 23.4, and 34.7 mm^2, respectively. Therefore, the longer fibers resulted in the lower elongation at break in the PDMS composites.

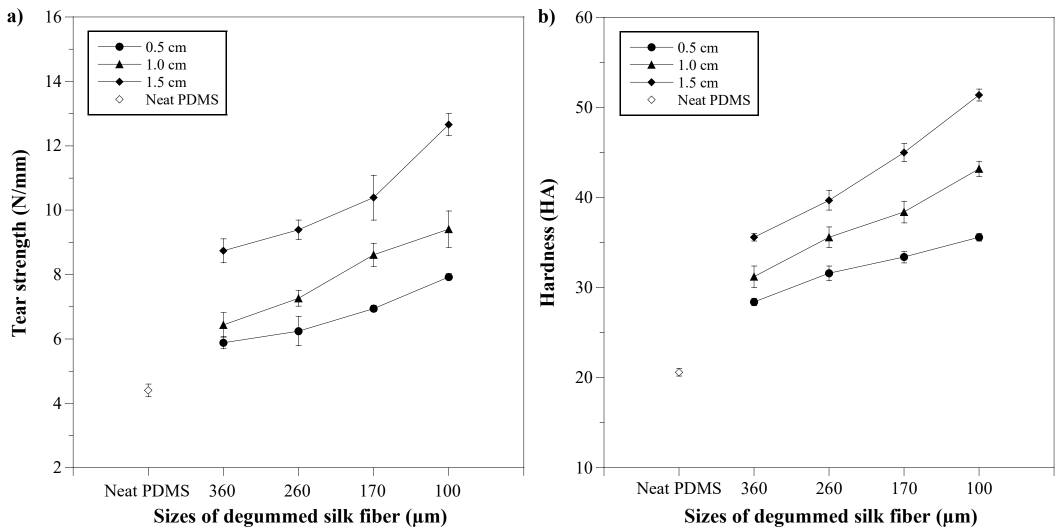


Figure 8. a) Tear strength and b) hardness of PDMS/DSFs having different fibers length and diameter (360, 260, 170, 100 µm) at constant fibers content of 1 phr.

3.3.2. Tear tests and Hardness

The tear strength of PDMS/DSFs having different fibers length and diameter at constant fiber content of 1 phr is illustrated in Figure 8a. It was observed that improvement in tear strength of PDMS/DSFs composites (360, 260, 170, 100 µm) is achieved by increasing the fibers length. This is due to the higher degree of surface area of long fibers which contributed to higher tear resistance. In addition, increasing the fibers length, the fiber bridging is also affected. The fiber bridging is formed, when the crack is initiated demonstrating that the fibers are linked each other and the matrix at the crack site, which means that longer fibers show higher tendency to cause a fiber bridging effect.[53]

The hardness of PDMS/DSFs with different fibers length was determined as well. The hardness increased with the increase of fibers length, which also affects the modulus of the composites.

3.3.3 Suture retention tests

The suture retention test measures the tearing resistance at the point of suture, where tension forces are applied. The suture retention strength is determined at the peak force during suture pullout.[30] The suture retention strength of PDMS/DSFs composites (360, 260, 170, or 100 μm) having different fibers length and diameter at constant fiber content of 1 phr was investigated. The results obtained are shown in Figure 9. The suture retention strength value of the neat PDMS was observed to be 2.6 N and increased significantly with an incorporation of DSFs. The results also showed a developing trend, for the fibers with smaller diameter. This is the same behavior as that shown in Figure 3 and explained in the corresponding text. Moreover, the suture retention tended to increase with increasing the fibers length, because of higher surface area between fibers and PDMS matrix. This helps to resist the pullout force of the suture wire. It is worth noting, that 1.5 cm length of the 170 or 100 μm resulted in too high hardness so the suture wire was worn out by specimen and the suture retention strength could not be measured (Figure 9). Moreover, the hardness of the composites with 170 or 100 μm fiber size is 45.0 ± 1.0 and 51.4 ± 0.7 HA, respectively, were much higher than that of human skin. Therefore, these composites were not suitable for mimicking the artificial hand skin.

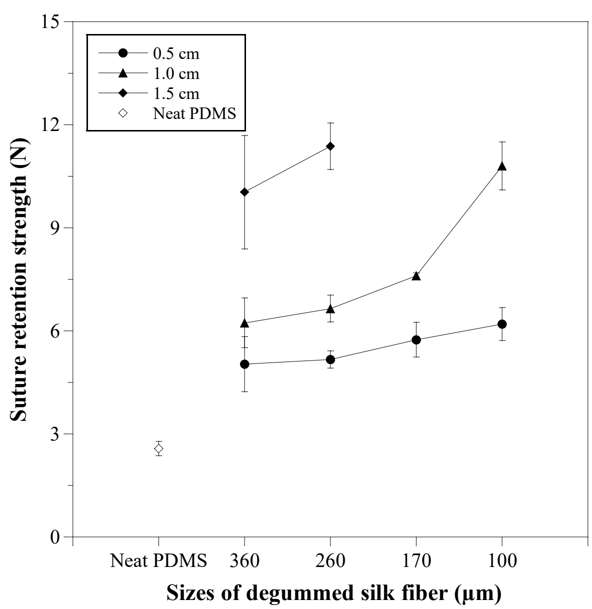


Figure 9. Suture retention strength of PDMS/DSFs composites having different fibers length and diameter (360, 260, 170, 100 μm) at constant fiber content of 1 phr.

The results displayed in Section 3.2 and 3.3, showed that 170 µm fibers size is the most suitable to be further developed as a hand surgery model for surgical training. The 170 µm size indicated greatly enhanced mechanical properties in terms of tear strength and suture retention. Thus, this fiber was used to study the effects of fibers content on mechanical properties, in order to find an optimum content for the artificial hand model.

3.4 Effects of Fiber Content of PDMS/DSFs composites on Mechanical Properties

Figure 10a shows the mechanical strength of the composites having different fibers content. It is seen that their tensile strength tended to increase with increasing fibers content due to higher number of interfaces serving more capacity load when receiving stress. In the same manner, the modulus of the composites was enhanced with fibers content, as shown in Figure 10b. This is mainly due to the high stiffness of silk fibers. In contrast, the elongation at break of composites decreased with increasing fibers content as illustrated in Figure 10c; this is due to the larger fibers volume that caused fiber clumping in PDMS composite.

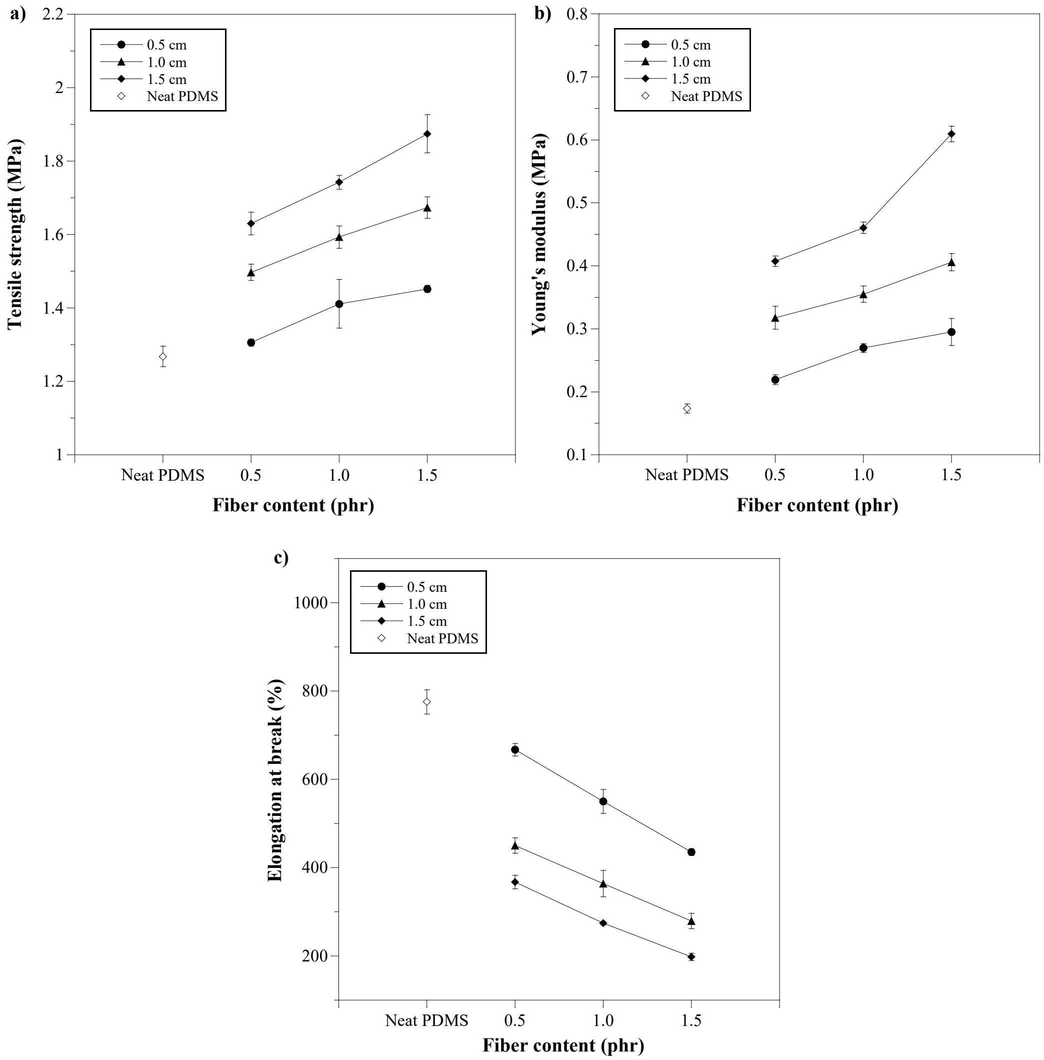


Figure 10. a) Tensile strength, b) Young's modulus, and c) elongation at break of PDMS/DSFs with size 170 µm fibers and different fibers content.

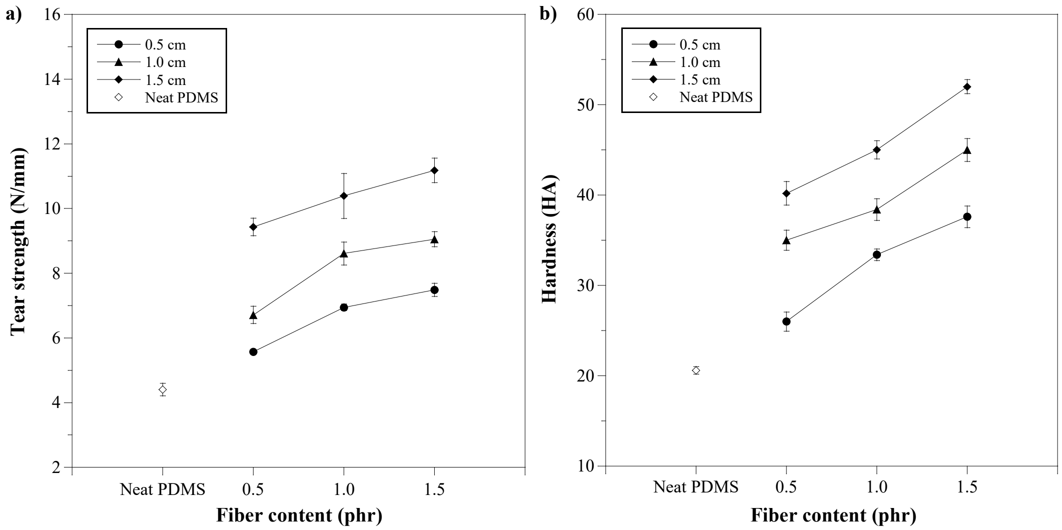


Figure 11. a) Tear strength and b) hardness PDMS/DSFs composites with 170 µm fiber size having different fiber content.

As displayed in Figure 11, the tear strength and the hardness of composites was increased with higher fibers content. This is due to the same reasons as the results shown in Figure 4.

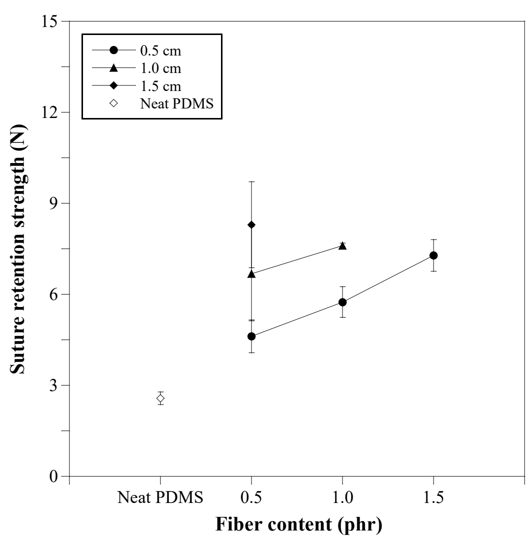


Figure 12. Suture retention strength composites with 170 µm fibers size having different fibers content.

In the same manner, the suture retention strength of the neat PDMS was increased with an addition of DSFs, as illustrated in Figure 12. Moreover, at higher fibers content, the suture retention strength of composites was increased remarkably. The increasing fibers volume indicated that the suture point would be more resistant to cracking. Nevertheless, the results showed the same behavior, as in Figure 9 for the suture wire's inability to penetrate the composite. Furthermore, the results showed large error bars at fibers content of 0.5 phr. This was because too few fibers were used leading to an uneven dispersion in the matrix. As a result, it is unable to resist cracking when the tearing force of the suture wire is applied.

3.5 Contact Angle of PDMS and DSFs

The contact angle measurements on natural fibers were conducted according to the procedure reported in Schellbach *et al.* (2016),[31] Table 1 show the measurement of d and h values and calculated contact angle. The result showed that the contact angles of water and PDMS on DSFs were 80.54 ± 4.0° and 35.40 ± 3.0°, respectively, indicating that PDMS has more compatibility with DSFs compared to water. Moreover, the DSFs showed hydrophobicity than other natural fibers for example hemp, curaua, fique, jute, piassava, pineapple, ramie and sisal in which they were reported to have water contact value to be in the range of 40-50°.[31] This significant result explained the greater compatibility between PDMS and DSFs resulting in the enhancement of the mechanical composite performance.

Table 1. Contact angle measurement between DSFs and water or PDMS.

Type of solution	d value	h value	Contact angle (°)
PDMS	23.3 ± 2.4	6.0 ± 0.8	35.4 ± 3.4
Water	26.3 ± 7.5	1.0 ± 0.2	80.5 ± 4.6

3.6 Rheology study of PDMS prepolymer containing DSFs having different fibers content and a fixed fibers diameter (170 µm) and fibers length (1.5 cm).

The rheology test examines the viscosity and flow phenomena of the uncured prepolymer PDMS/DSFs; viscosity is an important property for taking into consideration of processability. The rheological responses are related to the final form of the application in the manufacturing process. The steady shear viscosity of uncured PDMS/DSFs samples having different fibers content at a fixed fiber size of 170 µm and fiber length of 1.5 cm is illustrated in Figure 13. The viscosity tends to increase with increasing the fibers content at low shear rates. This could be caused by fibers in the matrix undergoing elastic fiber interlocking. Because of their elasticity, the fibers become locked together and form a network, which can be easily destroyed, as the shear rate increases, which results in decreasing viscosity at high shear rates.[54] These results indicated the samples presented a pronounced shear-thinning behavior, which was in agreement to the report by Setua.[55]

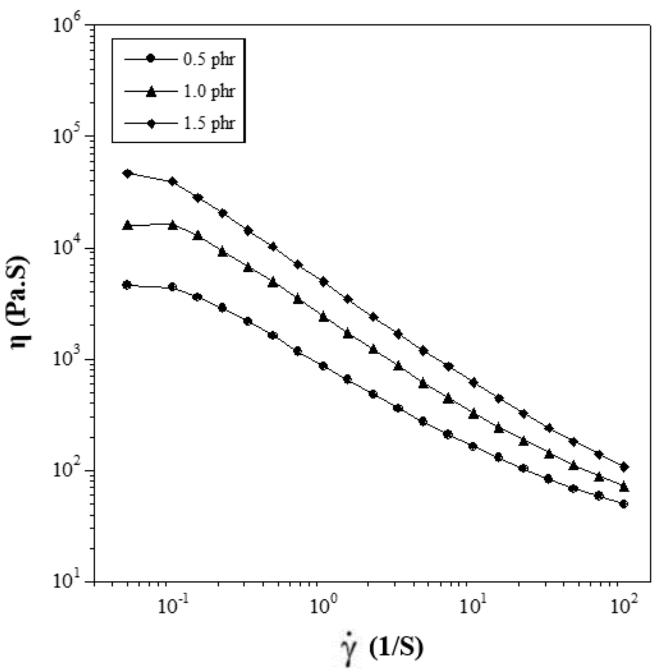


Figure 13. Steady shear viscosity of compound with fibers size 170 µm having different fibers content.

3.7 Surgical Artificial Hand Skin Evaluation

For development of artificial hand skin for suture practicing, the mechanical properties must be close to that of human skin, i.e., tensile strength of 1-2.5 MPa and hardness of 30-40 HA for the best practicing experience.[40,41,47] In this work, it was found that 170 µm size DSFs, fibers length of 1 cm, and fibers content of 1 phr was the most suitable. The PDMS/DSFs composites with 100, 170, and 260 µm fibers size were suture tested. The suture wire was cut by the composite with 100 µm size fibers due to too high hardness of the composite sample reinforced with 100 µm size DSFs, while the composite with 260 µm size DSFs showed a low value in suture retention strength. The composite with 170 µm size DSFs had high efficiency for surgical practice, which is consistent with the suture retention strength previously tested in Figure 12.

In conclusion, the composite with 170 µm size DSFs, with fibers length of 1 cm, and fibers content of 1 phr was selected to produce artificial hand skin with a thumb disability for surgical practice as illustrated in Figure 14a. The results suggested that the artificial hand skin was able to resist the tearing from suture wire and the material after surgical is shown in Figure 14b. Therefore, the material has a high potential to be used as the artificial hand skin for surgical practice for the best practicing experience.

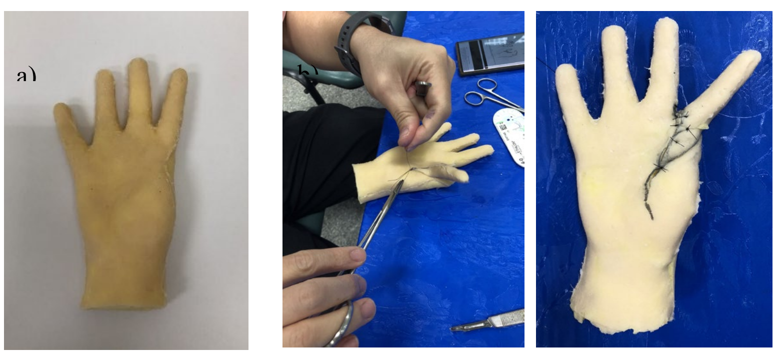


Figure 14. a) Artificial hand skin and b) surgical artificial hand skin practice.

4. Conclusions

A PDMS/DSFs composite was successfully developed for use as artificial hand skin. The mechanical properties of PDMS/DSFs were increased with an incorporation of DSFs. The results revealed that the fibers diameter is an important factor influencing the fibers contact area in the PDMS matrix; DSFs with smaller diameter more greatly enhance the mechanical properties of composites. Furthermore, the fibers length and fibers content are variables that both affect the mechanical properties, an increase in length and content results in improvement of mechanical properties. On the other hand, the addition of reinforcing fibers noticeably decreases the elongation at break of composites. Moreover, according to the SEM micrographs and contact angle measurements, the DSFs have good adhesion to PDMS matrix and a relatively uniform dispersion within the PDMS composite. Finally, it was found that the tensile strength, hardness, and suture retention strength obtained for composite with fibers size of 170 µm, fibers length of 1 cm, and fibers content of 1 phr is suitable to mimic real human hand skin for the best surgical training experience.

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Author contributions

Wongsalam, T. conducted experiment and prepared draft manuscript, Okhawilai, M. designed the experimental setup, analyzed data and edited final manuscript, Charoensuk, K. edited draft manuscript, Luangjarmekorn, P. conducted experiment, Karagiannidis, P. edited final manuscript, Boonnao, N. edited draft manuscript, Rimdusit, S. supervised the project and edited draft manuscript. All authors contributed to the discussion and reviewed the manuscript.

Competing interests

The authors declare no competing interests.

Data Availability Statement

All data generated or analyzed during this study are included in this published article.

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