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Research paper

Comparing friction of additively manufactured materials with animal blood vessels

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ABSTRACT

The replication of blood vessels for training and research purposes is possible with the help of additively manufactured (AM) models. However, a meaningful evaluation of the quality of the haptics, here concentrating on friction characteristics, of additively manufactured blood vessel models compared to human vessels is difficult and often only based on subjective assessments. To enable an objective comparison of friction of different AM materials, tests were performed in which a braided stent was pulled through straight test tubes. The force required to do so was measured. The same test setup was used to examine animal blood vessels so that these results could be compared with the findings of the AM materials. In addition, physicians were asked for their assessment of the haptics concerning friction of different materials. Summarizing the results, for the tested Formlabs materials *Flexible 80A* and *Elastic 50A, i*t can be stated that *Flexible 80A* is strongly recommended for the replication of blood vessels - even though it is comparatively smooth. The *Elastic 50A* should only be used for training with increased difficulty since the models are stickier and a flipping of instruments is possible. Coating the materials only involve effort that is not reflected in the benefits.

1. Introduction

To conduct training and research concerning vessel diseases, true-to-scale hollow vessel models are integrated into a medical simulator called HANNES (see Fig. 1) [1–3]. Thus, not only animal models can be replaced, but also the training and research frequency as well as reproducibility can be increased, which contributes significantly to an improvement of the skills of the physicians, and also to a sustainable and secured development of medical instruments. Due to the great geometrical freedom, the replication of human blood vessels is realized through additive manufacturing (AM) [4]. Patient-original and individual vascular models can be manufactured economically in small quantities [5,6]. Thus, vessel models, such as carotid, aneurysm or stenosis models, can be manufactured and integrated in the medical simulator [4,6].

In order to make research and training with AM models possible and desirable, it is essential that the interaction of the AM models with medical instruments behaves as close as possible to that in human blood vessels, offering the medical professional a realistic feeling during training and research activities. Only then, a medical simulator can replace human or animal experiments and provide reliable and realistic results and evaluations. Publications on the use of additively manufactured models for treatment training or in the context of research and further development of medical devices hardly address the problem of material selection for realistic haptics. Rather, they focus on the presentation of the development process or application of the developed models (e.g. [7–11]). The material selection is hardly ever explained or discussed in detail, often only the final selection is presented. The research question underlying this paper is therefore as follows.

> *What AM material should blood vessel models be made of to provide a realistic haptic for a physician (when evaluating personalized flow diverters)?*

One aspect to evaluate haptics is to investigate friction characteristics [12]. The aim of this paper is therefore to explain the material selection by presenting results of friction tests with different AM materials, post-processing methods and coatings in order to derive a recommendation for AM materials for its use in the field of endovascular training and research. Friction is a part of tribology. Tribology in general deals with the movement of one body over another [13]. A distinction can be made between kinetic friction, here especially sliding friction, and static friction [14]. The former considers the force required to move one body consistently over another, while the latter focuses on the breakaway force at the beginning of a motion, which is usually greater than the frictional force at the constant motion due to the adhesion of the materials [15]. If the static friction is noticeably

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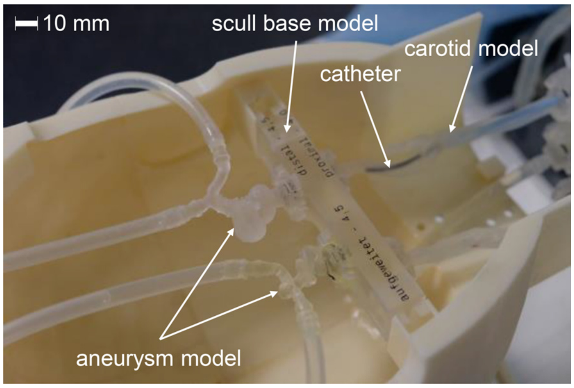


Fig. 1. Additively manufactured aneurysm models of the medical simulator HANNES.

greater than the sliding friction, an unstable friction behavior can occur in which rest (stick) and sliding (slip) alternate. This behavior is referred to as stick-slip movement [16]. The degree of interaction between two materials is strongly material-dependent [17]. To ensure that the physicians do not have to apply too much force when moving instruments in the blood vessel during minimal invasive treatments and possibly injure or, in the worst case, puncture the blood vessel, it is very important that the friction between the instruments and the blood vessel is minimized [18]. To keep the frictional forces of medical instruments as low as possible and to avoid stick-slip effects, for example hydrophilic coatings are applied [18].

In our case, however, the medical instrument should not be changed, but the AM material should be selected in such a way that the interaction between a medical instrument and the AM model is similar to that between a medical instrument and the human or animal vessel. Of particular interest in the present case is the friction pairing AM material - flow diverter. A flow diverter is a medical implant used for the treatment of intracranial vascular aneurysms [19,20]. The braided stent is inserted into the vessel system using a catheter system and then released in the diseased vessel, where it expands with a certain radial force due to the shape memory alloy. With the aid of the HANNES medical simulator, it is now being investigated to what extent personalized flow diverters influence the success of a treatment [2,21]. It is of great importance that the behavior of the flow diverter, e.g. with regard to migration i.e. slipping in the vessel (also see [22]), is very close to reality during and after implantation. Concerning AM models, this article focuses on the printing process of steriolithography [23]. Using this process, models are fabricated layer by layer from a liquid, synthetic resin by selective curing with UV light [23]. Support structures to enable the printing process are made of the same material as the part itself and can be removed mechanically after the printed part has been washed and cured [23]. For small structures, internal support structures can be avoided; the support structures provided on the outside of the part are often sufficient [6].

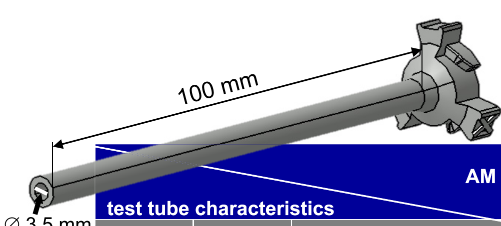
Before the results of the tests are presented in Section 3, the test samples and the test setup with the test procedure are introduced in Section 2. The results are then discussed in Section 4 before a conclusion is drawn in Section 5.

2. Material and methods

In this chapter, the research method is presented. First, the materials to be investigated are introduced. Then, the test setup and the test procedure are described, before the analysis and interpretation of the measured data is explained.

2.1. Test samples

In this paper, UV-curable resins by *Formlabs* are investigated. These are applied in a *Form 2* or *Form 3* printer of *Formlabs* and test tubes with a constant inner diameter of 3.5 mm, a length of 100 mm plus adapter and a wall thickness of 1 mm are produced (see also Fig. 2 or [24]). An adapter was designed at one end of the test tubes to ensure positioning and fixation in the test setup. Since the surfaces of the later aneurysm model are not parallel and aligned very differently in the build space of the printer during the printing process, an ideal alignment of the test tubes (e.g. horizontally or vertically) would not be helpful for the study and would falsify the result. Therefore, since a printing orientation with an angle of 10–20° is recommended by Formlabs to increase the printing success rate, a slightly diagonal orientation (~20°) of the tubes in the printing room is chosen with the adapter pointing upwards. The printing orientation is the same for all test tubes. Flexible materials were investigated, with



- transparent
- shore hardness 50 A
- stress at 100% elongation 1.6 MPa
- ultimate tensile strength ~ 2.2 MPa

- transparent
- shore hardness 80 A
- stress at 100% elongation 6.3 MPa
- ultimate tensile strength ~ 8.9 MPa

- black (predecessor of Flexible 80A)
- shore hardness 80-85 A
- ultimate tensile strength ~ 8 MPa

AM Material / test tube characteristics			Elastic 50A layer thickness 0,1mm	Flexible 80A layer thickness 0,05mm	Flexible layer thickness variant
washing	curing	specialty	name	name	name
20 min	20 min	none	Elastic50A	Flexible80A	-
20 min	20 min	printed more than 1 month ago	Elastic50A_old	Flexible80A_old	-
20 min	4h	long curing	Elastic50A_C	Flexible80A_C	-
40 min	20 min	long washing	Elastic50A_W	Flexible80A_W	-
40 min	4h	long washing and long curing	Elastic50A_W+C	Flexible80A_W+C	-
20 min	20 min	coating with OKS 1301	Elastic50A_OKS	Flexible80A_OKS	-
20 min	20 min	coating with teflon spray	Elastic50A_TS	Flexible80A_TS	-
20 min	20 min	coating with silicon spray	Elastic50A_SS	Flexible80A_SS	-
20 min	20 min	layer thickness of printing 0.05mm	-	-	Flexible_0.05
20 min	20 min	layer thickness of printing 0.1mm	-	-	Flexible_0.1

Fig. 2. Overview of the tested materials and their variations, including material characteristics provided by the material manufacturer.

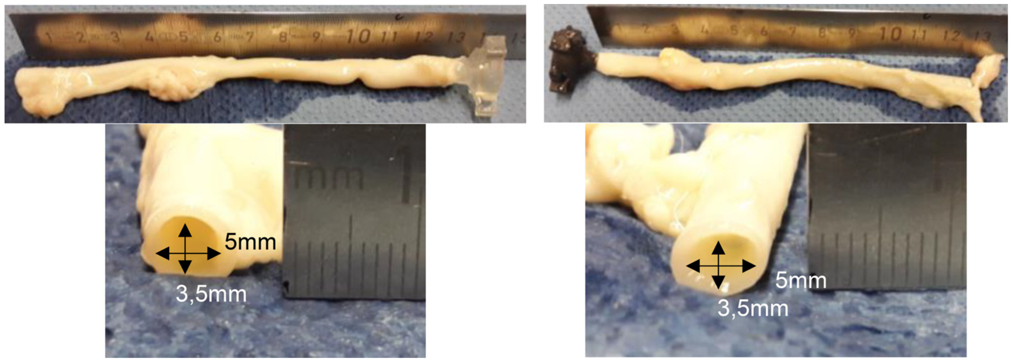


Fig. 3. Size of the examined sheep vessels.

the focus on flexible and transparent materials of *Formlabs* (Elastic 50A, Flexible 80A and Flexible). This is because, as in the human blood vessel, a flexible behavior of the model is to be achieved. The advantage of transparency is that optical systems can be used as an alternative to X-rays and, unlike in the patient, the behavior of the medical instruments can be observed optically from the outside. A further requirement for a suitable material was that it could be printed directly next to HANNES using the existing Formlabs printer. It should also be possible to apply coatings and the like easily and independently without the need for external help. Fig. 2. lists the materials with coatings or modified post-processing that were examined in the series of tests. Material characteristics of the pure materials out of the datasheet of the resin manufacturer are added [25].

A long washing time promises to reduce the stickiness of AM models, as resin residues are better washed out. A longer curing time is being investigated, as excessive curing of the model may also reduce stickiness due to resin residues. The same applies for models, which are printed a long time ago. Transparent coatings, which are known to reduce friction between objects, were chosen (teflon spray, silicone spray). In addition, a water-resistant OKS coating is chosen because the aneurysm models will later be flushed with water, for example, during a training session. All configurations are compared to a not modified AM model manufactured according to manufactures specifications. Furthermore, the influence of the printing thickness was investigated the material *Flexible*. Here, different layer thicknesses can be chosen while preparing the model for the printing process.

In addition, sheep arteries and pig vessels were examined (see Fig. 3). These were removed from a dead animal, stored in NaCl solution and examined in the test setup within a few hours. Adapters were glued to the vessels with superglue to allow positioning in the test setup. The sheep vessels were approximately 2–3 cm longer than the test tubes. In addition, it has to be remarked that the diameter varies along the course of the vessel.

2.2. Test setup and test procedure

The development of the test setup is presented by Kuhl et al. [24]. The improved test setup is shown in Fig. 4. To sum up, a 20 mm out of the introducer released flow diverter of size 4 × 60 mm is pulled through the test tubes. The pulling movement is realized by a universal testing machine. The force is measured by an S-shape force sensor of type KD24s +/−10 N, which has an accuracy of 0.1%. The sampling rate is 100 Hz. The measured data is recorded using the Diadem software. The experiment takes place in a water bath tempered to about 35 °C, so that the medical instruments behave as in the human body (requires >30 °C).

Two different test procedures are used: First, the flow diverter is pulled through the tube with a constant speed of 200 mm/min to

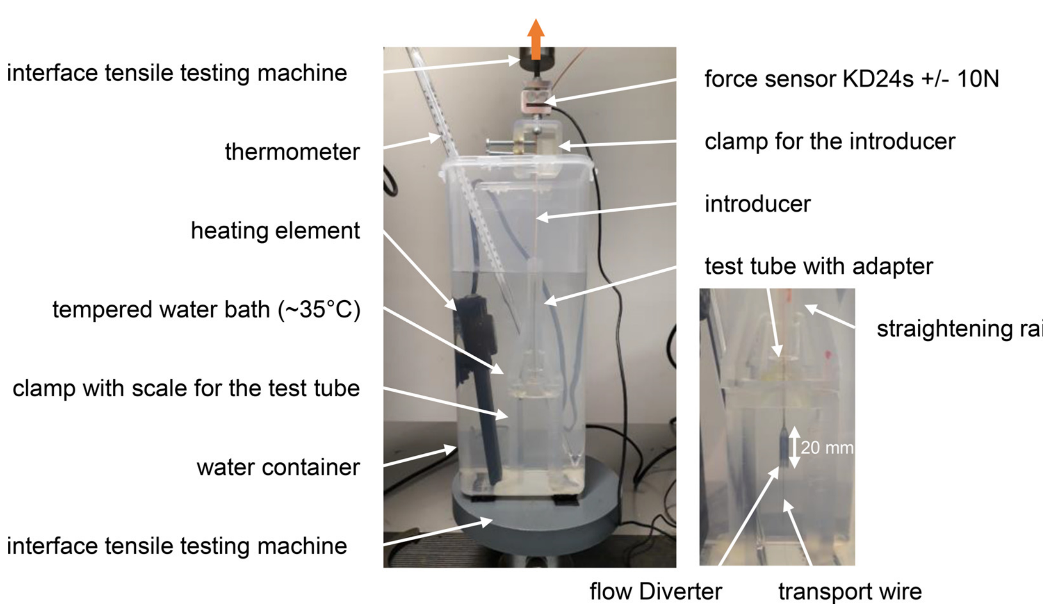


Fig. 4. Test setup for evaluating the static and sliding friction forces of different materials.

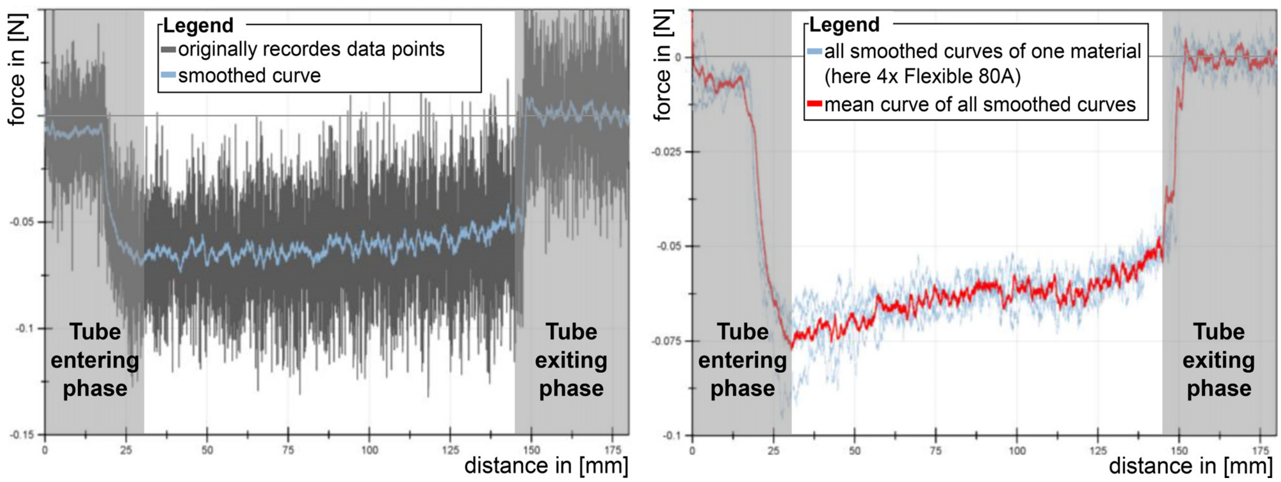


Fig. 5. Clarification of smoothing (left) and definition of mean curves (right) for evaluating the measured data.

investigate the sliding friction force. In addition, tests are run in which the pulling movement is stopped three times while the flow diverter is inside the tube, and then restarted to investigate the static friction force. At least two identical tubes of each material or coating were tested, each at least twice.

2.3. Analysis and interpretation of the measured data

In order to evaluate and compare the measured data, they were processed the same way. As already described by Kuhl et al. [24], the tube entering phase at the beginning (0 to 30 mm) and the tube exiting phase (after 145 mm) are not taken into account, since the flow diverter is not completely inside the test tube. When the flow diverter is pulled into the tube, it elongates, so that even though it only emerges 20 mm from the introducer, a distance of up to 30 mm must usually be covered until it is completely inside the tube. In addition, the measured values were smoothed to dampen measurement noise and errors. Here, the +/−16 neighboring values plus the value itself were used to form the mean value for a value (see Fig. 5, left).

The sliding friction forces are evaluated qualitatively in graphical form and quantitatively in tabular form. Since a material configuration was tested repeatedly, the different curves of a pairing were combined into a mean curve (see Fig. 5, right). In this way, different material pairings can be compared with each other graphically more easily while considering consider multiple tests at the same time. The graphical evaluation serves in particular to assess the stick-slip behavior and the rough order of magnitude of the forces. Having theoretical friction basics in mind, the stick-slip effect can be concluded from the oscillating force curve. Since the lateral force F_N as well as the coefficient of sliding friction $\mu_{sliding}$ can be assumed to be constant - since the tube and flow diverter can be assumed to be constant - the oscillation of the measured force values F_F can be explained by an alternation between sliding and sticking modes, switching between $\mu_{sliding}$ and μ_{static} (since $F_F = F_N{}^*\mu$). This statement is supported by the visual observations of the stent movement in the transparent tubes. For the quantitative evaluation, to compare the sliding friction forces, an average value of the data between 60 mm and 130 mm was determined for each test. Additionally, the average value over all tests of one material configuration is formed.

In the analysis and interpretation of the static friction forces, the force transition when starting up after all three stops is examined. The magnitude of the force transition is read out manually from the curves and data tables for each stop and each material pairing.

Finally, selected materials are also explored and evaluated subjectively by two physicians, who test different AM aneurysm models that are integrated in the simulation model HANNES, in random and unknown order, some models twice. After probing the aneurysm model, the interaction between medical instrument and AM model is assessed by the physician on a scalar from 1=very unrealistic to 10=very realistic feeling. In this way, the subjective assessment can be contrasted with the interpretation of the results from the objective measurement.

3. Results

Fig. 6 shows the graphical evaluation of the mean curves of the materials *Elastic 50A, Flexible 80A* and *Flexible* as well as the curve of the sheep vessel.

It can be seen that the curve of *Flexible* shows the smoothest curve, indicating no pronounced stick-slip effect, and the lowest dynamic frictional force values of all presented materials of Fig. 6. The *Flexible 80A* has slightly higher sliding frictional force values and a stronger oscillating curve. The *Elastic 50A*, however, shows in comparison a strong stick-slip effect, which could also be seen with the naked eye during the tests, and which is expressed by the highly oscillating curve in Fig. 6. In the case of the sheep vessel, due to the gluing of the adapter and the naturally uneven vessel diameter, the vessel is narrowed at the beginning (see Fig. 3, left), which shifts the entire curve of the measured sliding friction forces downward. The curve can only be evaluated from approximately 80 mm. Overall, the sheep curves oscillation is comparable with *Flexible 80A* and does not show a strong stick-slip effect. The magnitude of the measured force values, in contrast, is more in line with the *Elastic 50A* material.

The investigation of different post-processing steps has shown that the strong stick-slip effect of the *Elastic 50A* material can be reduced by a longer curing time since the force curve becomes significantly smoother. Similar applies when *Elastic 50A* is coated with silicone or teflon spray, but the coatings are washed out after a short time in the water bath. The water-resistant OKS is more resistant and shows the same effect on the stick-slip effect.

A longer curing time, though, is hardly noticeable with the material *Flexible 80A*. A difference in the already low force oscillation cannot be seen graphically. A more extended curing time also causes the material to lose elasticity and become significantly stiffer, which is considered by the medical experts to be more unrealistic. Other coatings or post-processing steps hardly reveal any difference in the force curve for *Flexible 80A*.

The quantitative evaluation with the average sliding friction forces of the materials per test are shown in Fig. 7. In a dark blue color, the lowest column per material shows the average sliding friction force across all tests. The results of different tests in the same test tube are painted in the same color. It can be seen that only

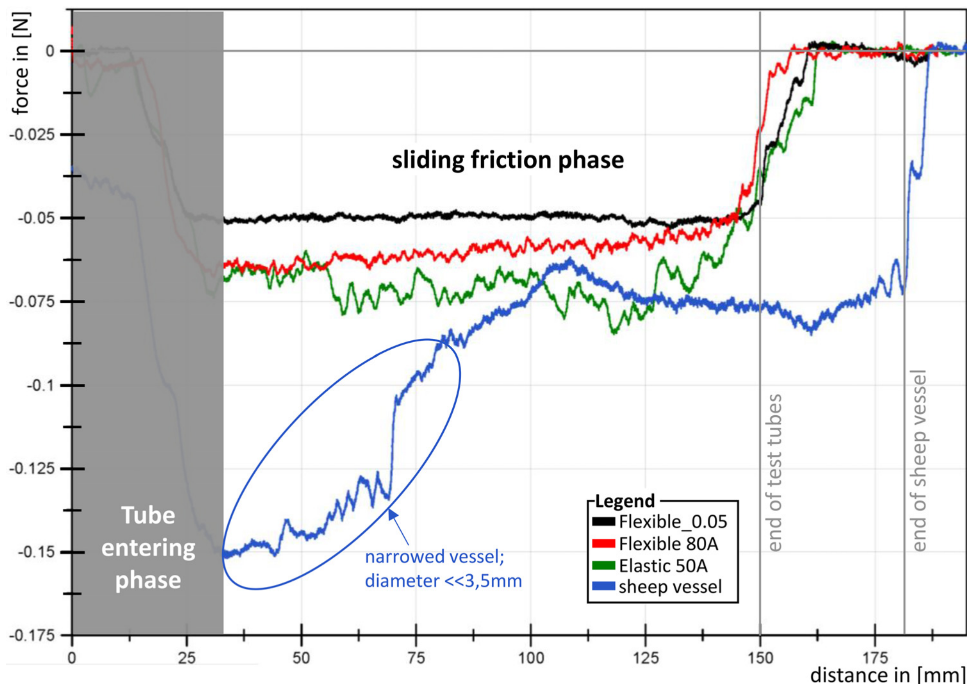


Fig. 6. Diagram of the mean curves of sliding friction forces of the materials *Flexible, Flexible 80A, Elastic 50A* and a sheep vessel.

minimal differences in the average sliding friction force can be measured between two tests in the same tube, confirming good repeatability of the test setup. Even between the test tubes of the same material setup, only minor differences occur.

The figure shows that the magnitude of the sliding friction forces of all materials is very similar and on the whole corresponds to those of the sheep vessels and the investigated pig vessel. In the case of *Elastic 50A*, the quantitative evaluation also shows the picture just described: the average sliding friction force is reduced by a changed post-processing or a coating. Old tubes and long cured tubes show more or less the same behavior. The coated test tubes also have a slightly reduced average sliding friction force. In the case of *Flexible 80A*, it is noticeable that all the sliding friction force magnitude of all the tests are almost the same. Coating or special post-processing after printing does not significantly change the force value and is therefore additional workload without effect. The layer thickness during printing also does not seem to have a strong influence, as shown by the tests with *Flexible*. In the case of the sheep vessels, it is noticeable that the frictional force decreases from one test run to the next. A possible explanation for this is that plaque and blood residues are increasingly scraped off with each trial, making the vessels smoother and smoother, thus reducing the sliding frictional force.

An investigation of the static frictional forces does not yield any reliable results. Due to the magnitude of the forces, the measurement noise is large, so that with reasonable workload no accurate distinction could be made between the peak of static friction and the subsequent sliding friction. Therefore, no qualitative and meaningful evaluation of the measured force values is performed. Nevertheless, the evaluation of the sliding friction forces is sufficient for this investigation. If static friction forces would be important, optimizations with regard to the test setup and the test procedure can to be considered and implemented (e.g. higher sampling rates), the data analysis can be modified and adapted, or a more accurate sensor can be used.

The subjective assessment by physicians shows that *Elastic 50A* comes very close to the haptics concerning friction in human vessels, but it still seems a little too sticky since, for example, the guide wire suddenly jumps on and unexpectedly after great force are applied. In *Flexible 80A*, on the other hand, the guide wire slips through the vessel a bit too easily. Differences between models with and without coating of one material could not really be identified.

4. Discussion

Overall, it can be summarized that the replication of blood vessels for medical models using additively manufactured materials is possible. Different materials are suitable to different extents. It should be kept in mind that the haptics with regard to the friction characteristics in humans are not uniform either, and plaque deposits and other typical irregularities on the blood vessel wall also result in large differences in the interaction with medical instruments and implants. Also, friction alone does not make haptics, but in this specific application it represents a major part.

The magnitude of the sliding friction forces is comparable to the sheep vessel in all tested and evaluated material configurations. The *Elastic 50A* material comes closest to the sliding friction force values of the sheep vessel. However, the stick-slip effect of the material is too pronounced. The curve of the *Flexible 80A* material agrees much better with that of the sheep vessel, but the sliding friction force magnitude of the material is somewhat too low. Coatings or special post-processing only show a slight effect with *Elastic 50A*, but the result hardly justifies the work. When evaluating the tests, it must be clearly remembered that force differences of 0.01 N can hardly be resolved due to the accuracy of the force sensor, and the results must therefore be evaluated critically. With the magnitude of the forces, the results are very sensible to the measurement noise and disturbance factors, such as a slight vibration of the pulling machine. These cannot be completely ruled out, although all variables were attempted to be kept constant. Also, the smoothing and the formation of the means should be kept in mind. Since the determination of a magnitude of a friction force is clearly more difficult [26], the focus in this paper is the comparison of materials in the defined test setup.

Based on the results, together with the subjective assessment of the physicians, the following recommendations can be made: Simple probing of cerebral vessels can be well replicated with the *Flexible 80A* material - if, for example, learning the basic procedure and handling of instruments and implants is the main focus when using the

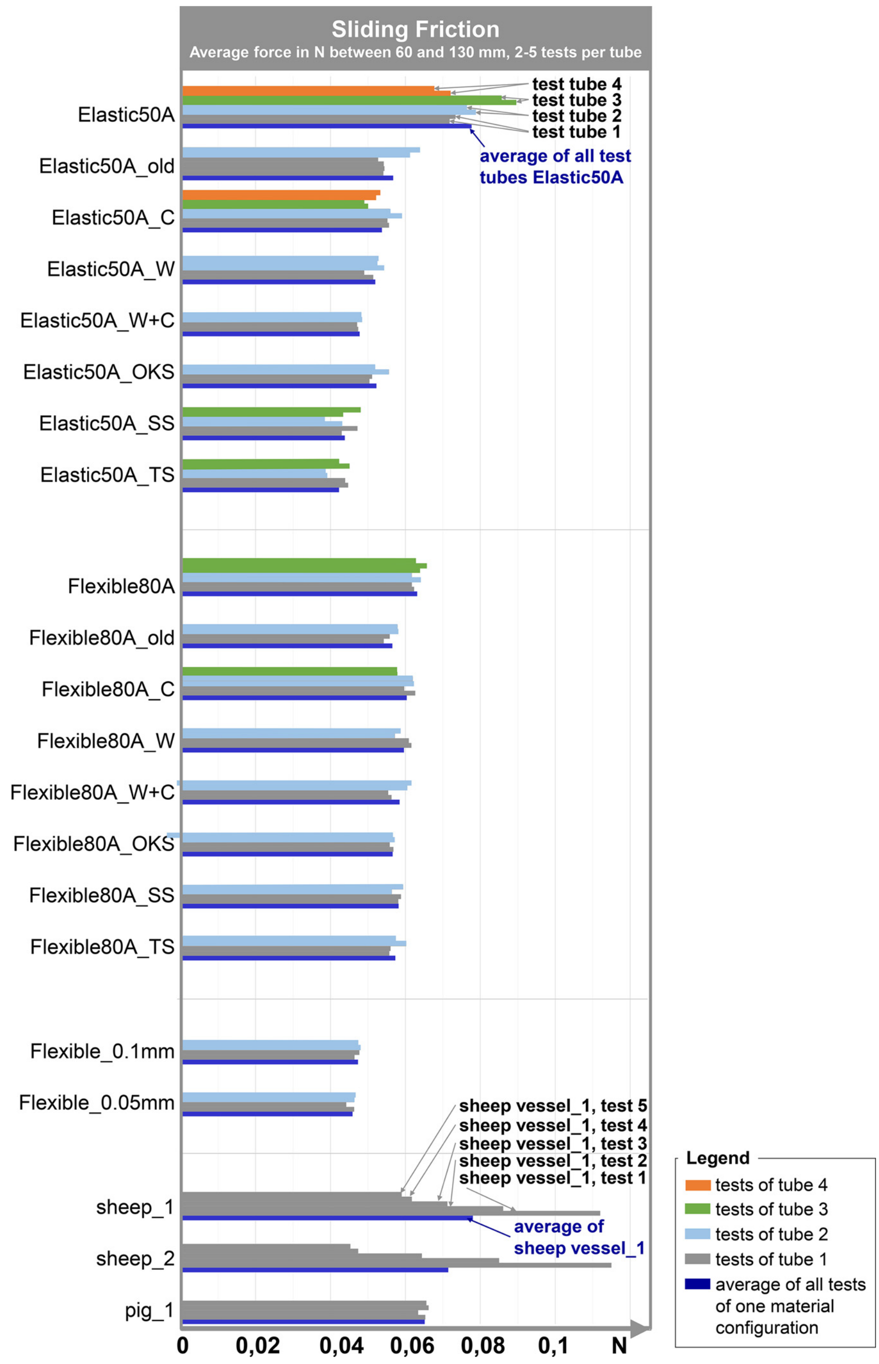


Fig. 7. Average sliding friction force of the analyzed material configurations.

model. The material is well suited for replicating diseased vessels, but is somewhat too smooth to accurately replicate the feel of uneven and diseased vessels. However, *Flexible 80A* is strongly recommended for the arterial pathway from the access to the diseased vessel. The *Flexible*, meanwhile, has the least realistic properties, as it is too smooth to replicate (especially diseased) vessels. The *Elastic 50A* material can be used for the training of difficult and uneven vessels. It remains to be considered that the material in interaction with a medical instrument makes a sticky impression and can provide a false haptic to the medical practitioner. These conclusions are based on the study of the AM material - flow diverter friction pairing. For the investigation of personalized flow diverters, the *Flexible 80A* material is chosen.

It cannot be excluded that the behavior of organic vessels from sheep or pigs changes after removal from the body. An attempt was made to counteract this with rapid testing of the vessels after the time of extraction. It also remains to be noted that the true arteries do not have a constant internal diameter. The vessels were selected in such a way that the inner diameter roughly corresponds to that of the test tubes.

5. Conclusion

This article presents the sliding friction behavior of additively manufactured test tubes and compares it with that of animal vessels. After interpretation, it can be stated that coating the Formlabs materials *Flexible 80A* and *Elastic 50A* involves effort that is not reflected in the benefits. The *Flexible 80A* is strongly recommended for the replication of blood vessels - even though it is comparatively smooth. The *Elastic 50A* should only be used for training with increased difficulty. The validity for other friction pairings can be investigated in the future. The same test setup can be used for this purpose. Overall, this study contributes to a greater integration of additively manufactured models into the daily training and research routine of physicians, thus helping to improve healthcare through repeatable and more frequent training and testing.

Declaration of interests

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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