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Comparison of the friction behaviour of occluded human skin and synthetic skin in dry and moist conditions

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Abstract

The goal of this work was to assess the suitability of a commercial synthetic skin to simulate occluded human skin friction behaviour in dry and moist skin conditions and under different applied surface pressures, with the view to using this material as a tribological test-bed for healthcare and personal care devices that are in direct contact with the skin during use. A flat rotating ring friction measurement device, in which one part of the skin surface is continuously covered (i.e. occluded), was used to compare the friction behaviour of human skin and the synthetic skin in controlled nominally dry and nominally moist skin conditions. Three loading levels were tested, simulating light, medium and high skin pressures typical of many lifestyle- and personal health-related applications. The results showed that the friction behaviour of the synthetic skin tested here was notably different to that of human skin in vivo in terms of the effects of skin hydration, sliding time and applied surface pressure. It is concluded that, for use as a tribological test-bed, the tested synthetic skin model does not provide an acceptable alternative to in vivo tests using human skin.

Keywords

biotribology, skin, artificial skin, friction

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INTRODUCTION

With devices that involve any form of physical interaction with human skin, the friction behaviour against the skin has an important though complex and not well-defined influence on the tactile perception of the device [1-3] and thus the user's experience. In many cases the skin friction behaviour also determines or affects the primary functional performance, such as where grip performance is integral to the functionality [4], examples being devices for the physically weak or disabled and sports equipment. Also many lifestyle- and personal health-related applications ranging from electric shaving to skin care devices [5], where the friction-influenced skin deformation plays an important role. Such devices may be used under different loading conditions, and often the skin surface is occluded by the device for much of the time of use. In all cases, during the development of the device there is a need to investigate and objectively quantify the skin friction behaviour in such a way that the results relate to the actual use conditions.

Measurement of the friction behaviour of human skin in vivo has a number of drawbacks: the measurements are often poorly reproducible due to person-to-person variability and involuntary human motions during testing, and the possibility of skin damage limits the severity of the conditions that can be applied. These disadvantages can hamper the interpretation of the results to such an extent that conclusions useful for product pre-development purposes are difficult to draw. For these reasons the use of synthetic skin substitutes with consistent and reproducible properties is desirable. Because skin-contacting devices may be used in different climates, it is important that the synthetic skin model is capable of representing human skin friction behaviour under different environmental and other usage conditions. In a humid climate the skin hydration increases and the friction can be dramatically higher than with dry skin conditions; an increase of up to a factor 10 in coefficient of friction is not unusual [6-8].

Many materials have been used in attempts to simulate the tribological behaviour of human skin in various conditions and against various materials [9-23]. Many of these studies have concerned the friction of skin against fabrics and textiles, often for the purpose of studying skin blistering and decubitus, or the study of cosmetics applied to the skin. There are relatively few publications that are concerned with the use of skin equivalents to simulate friction against healthcare and personal care devices. In particular, the commercially-available synthetic skin SynTissue™ from Syndaver™ Labs. [24], which has been developed specifically for use in the design verification and validation testing of medical devices, has received surprisingly little attention as regards tribological behaviour in the literature.

The goal of the work described here is to assess the suitability of the SynTissue™ synthetic skin model to simulate occluded human skin friction behaviour in dry and moist skin conditions and under different applied surface pressures, with the view to using this material as a tribological test-bed for skin-contact personal care devices. A flat rotating ring friction measurement device [6], in which one part of the skin surface is continuously covered, was used to compare the friction behaviour of human skin and the synthetic skin model in controlled nominally dry and nominally moist skin conditions. Three loading levels were tested, simulating light, medium and high skin pressures typical of many lifestyle- and personal health-related applications.

EXPERIMENTAL DETAILS

Skin preparation

The general characteristics of the human and synthetic skins tested are given in Table 1. The synthetic skin used was SynTissue™ from Syndaver™ Labs, with a thickness of 1.2 mm. Prior to testing, the human skin of the subject's left forearm was cleaned using a non-degreasing soap and dried with a non-lint towel. The skin was then left for approximately 30 minutes before testing to acclimatise in the pre-stabilised environmental conditions of a climatic room ("dry" condition: 21°C, 38% relative humidity, "moist" condition: 28°C, 85% relative humidity). To achieve the moist skin hydration condition, the forearm was additionally wrapped in transparent plastic cling-wrap for 10-15 minutes immediately prior to testing. The resulting occlusion of the skin led to the secretion of additional sweat and a high skin hydration.

The synthetic skin, with a thickness of 1.2 mm, was cleaned and stored in distilled water at room temperature. The synthetic skin samples used in all of the tests were cut from the same sheet of material in order to avoid sample-to-sample differences in composition, surface topography etc. For the "dry" test condition, the material was taken out of the distilled water prior to testing and excess water was squeezed out of the bulk material using a paper towel. The surface was then thoroughly dried with a fresh dry paper towel. For the "moist" test condition the material was not squeezed; only the excess surface moisture was removed from the surface by lightly dabbing the surface with a dry paper towel. Prior to the friction testing in the pre-stabilised "dry" or "moist" environmental conditions of the climatic chamber, the synthetic skin strip was placed on top of the human forearm, in the same area used for the human skin tests, and left to acclimatise for a few minutes before testing to achieve a surface temperature close to that of the human skin, see Figure 1c. The synthetic skin was left on top of the human forearm during the friction test.

The hydration condition and surface temperature of the human and synthetic skins were measured immediately prior to each friction test, see Figure 1 and Table 1. The average surface

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temperature was measured three times using an infra-red temperature meter (Hyelec MS6530, 2% accuracy). A Corneometer® CM 825 was used to measure the hydration level in five locations to obtain average values. The Corneometer® is based on a capacitance measurement and has a measurement depth of 10-20 µm [25]. For human skin, measuring the capacitance effectively provides an indirect measurement of the water content of the stratum corneum. This is because the dielectric constant of water is more than 10 times higher than that of the material of human skin, so that the capacitance of the water contained in the skin dominates the measurement [25]. The material used for the commercial SynTissue™ synthetic skin tested here is unknown. However, the dielectric constants of common polymers that have been used elsewhere in synthetic skin substitutes such as polyurethane elastomer (4-10) [26] and silicone rubber (3-10) [27] are significantly less than water at about 80. The dielectric constant of SynTissue™ is reported by the manufacturer to be validated against that of human skin [24]. This suggests that the capacitance of the water contained within the synthetic skin when hydrated may dominate the measurement as with human skin, and that the Corneometer® provides an indirect measurement of the water content of the uppermost 10-20µm of the synthetic skin. The Corneometer® gives a measurement value in Arbitrary Units (AU) ranging from 0 (extremely low) to 130 (extremely high). Commonly, a value <30 AU is considered to indicate very dry skin, 30-40 AU dry skin and >40 AU “normal skin” [25]. As a reference, normally hydrated stratum corneum has been shown to have an actual water content of 30-50% of its dry weight [28, 29].

Friction testing

The friction tests were carried out using a rotating flat steel ring which, when held flat against the

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skin in one position, slides unidirectionally with a circular motion on the skin. This results in torsional strain of the skin. The rotating ring was applied to human skin in vivo (volar forearm of one subject) and SynTissue™ synthetic skin under "dry" and "moist" skin hydration conditions. The bespoke rotating ring friction torque apparatus has been described elsewhere [6], see schematic in Figure 2. This gravity-compensated custom-built device enables friction measurements to be made by the test subject at any location on the body. A predetermined normal force is set by an adjustable spring tension and a predetermined rotational speed is set by adjustment of the motor voltage. The flat stainless steel test ring (outer diameter 12 mm, inner diameter 8 mm) is driven by a DC motor and is described in more detail in Table 2. An integrated torque transducer measures the friction torque directly between the skin and the rotating ring. The coefficient of friction (CoF) can then be calculated by dividing the measured friction torque by the product of normal force and ring radius.

The rotational speed of the device was calibrated using a stroboscope and set at 2.56 rev/s, which is equivalent to a sliding speed of 96 mm/s at the outer ring-probe diameter of 12 mm. Tests were carried out at three levels of normal force on the ring probe (0.31 N, 0.94 N and 1.56 N) for the "dry" and "moist" skin conditions described in section 2.1. The ring was cleaned before each test using isopropanol. A custom-made arm-rest with hand grip was used so that the subject could hold their arm securely during testing, minimising involuntary movements of the forearm or probe head. The rotating ring probe was pressed carefully on to the forearm of the test subject so as to occlude the skin surface during the friction measurement, see Figure 3.

Table 2 gives a summary of the test µconditions employed in the friction experiments. For each combination of surface pressure (3x) and environmental condition (2x), 4-5 repeat tests were carried out on different locations on the forearm or forearm with synthetic skin to obtain average values for the friction force behaviour over time. The test locations were chosen to be immediately adjacent to each other, occasionally with a small overlap. The skin was left for a few minutes between each repeat test to re-acclimatise. Note that with the synthetic skin, which was placed on top of the human forearm, the physical adhesion at the synthetic-human skin interface was sufficiently high to prevent any relative motion occurring at this interface during the test.

RESULTS

Friction behaviour for dry skin conditions

The coefficient of friction (CoF) depended on the sliding time and on the applied surface pressure. Characteristic plots of the friction behaviour as a function of sliding time for the human skin and synthetic skin material are shown in Figure 4 for the “light” and “high” surface pressures applied in the experiments. These plots show the CoF before, during and after placement of the rotating friction device on the skin. The average CoF results obtained under dry conditions for all surface pressures at the start and end of the test are given in Figures 5 and 6 for the human skin and synthetic skin respectively. For the results at the start of the test the values shown represent the CoF averaged over a period of 0.5 to 1s. The results at the end of the test represent the average CoF for the final 0.5 to 1s of testing. At least four repeat tests were carried out and the results averaged.

For the in vivo human skin at the “light” surface pressure of 5 kPa, the CoF remained relatively stable for the whole of the test duration. The average CoF was lower for the higher surface pressures of 15 and 25 kPa but in these cases increased during the test.

The CoF of the synthetic material is clearly much higher than that of the human skin, with values up to fifty times higher at the "light" 5 KPa surface pressure. At the "light" surface pressure of 5 kPa the synthetic skin showed an increase in friction coefficient during the test and compared to the value measured at the start, but at higher surface pressures the average CoF did not change significantly during the test period. As for the human skin, the CoF of the synthetic skin was lower at higher surface pressures. No stick-slip phenomena were observed during the tests.

Friction behaviour for moist skin conditions

Examples of the change in friction coefficient as a function of sliding time for the human and synthetic skins at "light" and "high" surface pressures are given in Figure 7 a-d. Additionally, Figure 8 shows the average CoF results obtained for the human and synthetic skins in the moist condition at the beginning of the test (averaged over a period of 0.5 – 1s) and at the end of the test (averaged over the final 0.5 to 1s of testing). The values shown represent averages from at least four independent tests.

At the "light" surface pressure of 5 kPa the friction of moist human skin was characterized by a gradual increase in CoF as a function of sliding time. This behaviour is illustrated in Figure 7a. The rate of increase varied substantially from test to test, leading to a relatively large standard deviation for the CoF at the end of the test, see Figure 8. At higher applied surface pressures the friction began to increase immediately from first application of the friction device on the human skin, see example in Figure 7b. In these cases, because of the rapidity of the friction increase, it was not possible to discern a meaningful value for the friction at the start of the test.

The CoF of the moist synthetic skin showed a large standard deviation between repeat tests, particularly at the lowest applied surface pressure of 5 kPa where both increases and decreases in friction with sliding time were observed. However, it is clear from the results given in Figure 8 that, for the “light” 5 kPa surface pressure, the average CoF of the moist synthetic skin at the start of the test was always substantially higher than that of moist human skin. At the “medium” applied surface pressure (15 kPa) the average CoF started at a lower value than for the “light” surface pressure, and again remained relatively stable throughout the test duration. At the test end the average CoF was much lower than that of the moist human skin at test end, see Figure 8.

At the “high” surface pressure of 25 kPa, the friction behaviour of the moist synthetic skin over time was characterised by a relatively low average CoF at the onset (lower than at “medium” applied surface pressure) that gradually decreased further to a constant value. This behaviour is illustrated in Figure 7d and is clearly different to that of moist human skin at the same surface pressure, see Figure 7b.

As with the dry conditions, with both human skin and synthetic skin, no stick-slip phenomena were observed during the tests.

Comparison between dry and moist conditions

Figure 9 compares the average CoF for the human and synthetic skins at the test end for dry and moist conditions. Note that it is not possible to compare values at the start of the test owing to the aforementioned rapid increase in CoF for the moist human skin at the “medium” (15 kPa) and “high” (25 kPa) surface pressure levels.

DISCUSSION

Effect of dry or moist skin conditions

The friction behaviour of the synthetic skin tested here was notably different to that of human skin in vivo. Compared to human skin, a much higher average CoF was observed for the synthetic skin at the start and end of the test in dry conditions, and, at least evidenced for the "light" surface pressure, at the start of the test in moist conditions. A further striking difference is that changing the conditions from dry to moist led to a large increase in the CoF with human skin but to a large decrease with the synthetic skin. The latter was observed for all applied surface pressures, see figure 9. In order to explain these observations it is necessary to consider the mechanism of skin friction and how the friction behaviour may be affected by structural differences between human skin and the synthetic skin.

There is much evidence suggesting that in most cases the friction of human skin is governed predominantly by an adhesion friction mechanism [6, 7, 30]. Capillary adhesion, through the formation of fluid menisci between the contacting surfaces, may in some cases contribute to the overall adhesion [31]. Notable exceptions where friction mechanisms other than adhesion play a more significant additional role are when the skin is very wet with a fluid film on the surface, where hydrodynamic lubrication can occur [7, 32- 34], and ridged-skin friction against very rough surfaces, where friction due to ploughing or deformation of the skin ridges by relatively hard roughness asperities can occur [35-38]. In the current work, there were no finger ridges present and the surface of the steel ring in contact with the human and synthetic skin was relatively smooth (R_a 0.29 µm), so that the ploughing/deformation friction mechanism is unlikely to be significant. Friction due to viscoelastic hysteresis as a result of changes in the strain of the skin tissue can in our case be considered negligible because of the test configuration used. Because the flat steel ring slides unidirectionally in a circular motion on the skin surface and no stick-slip behaviour was observed, the skin strain does not change direction during the test. The amount skin strain does increase or decrease due to changes in the friction coefficient resulting from other factors, but these changes occur relatively slowly over a period of seconds, so the losses in energy due to viscoelastic hysteresis are likely to be small.

With the adhesion friction mechanism, the adhesion friction force F_{adh} is given by the product of the interfacial shear strength τ and the real area of contact A_R:

$$F_{adh} = \tau \cdot A_R \qquad (1)$$

With human skin, interfacial shearing is thought to occur within the thin organic surface layer on the skin stratum corneum (SC) [6, 7, 30]. Depending on the skin condition, this surface layer can consist of an emulsion of varying amounts of water, epidermal and sebaceous lipids, sweat, loose corneocytes and other contaminants [39-42].

The synthetic skin does not have the complexity of human skin, for which the SC has a thickness in the range 10 to 40 µm and the viable epidermis, dermis and hypodermis layers are of varying thicknesses and have different mechanical properties [43]. It is also unlikely to have an organic surface layer equivalent to that on human skin. According to the manufacturer [24], the tensile modulus of the synthetic skin is validated against human skin. This suggests that the stiffness has been chosen to simulate the bulk of human skin, for which the dermis has an effective elastic modulus of 8-35 kPa, rather than the SC, which has a much higher elastic modulus of 10 – 1000 MPa depending on the hydration state [43]. On this basis, a hypothesis for the much higher average CoF of the synthetic skin compared to human skin is that the real area of contact at the interface is greater due to a lower effective elastic modulus than the SC on human skin. In the absence of a fluid film on the surface, the interfacial shear strength may also be higher than that of human skin with its complex organic surface layer.

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With human skin, it is well established that the stiffness of the SC decreases with increased skin hydration [44-49], and an increase in skin friction with increasing skin hydration has also been reported [6]. A decrease in skin stiffness leads to an increase in the real area of contact and hence greater adhesion friction with moist skin compared to dry skin.

The contrasting behaviour of the synthetic skin can be attributed to the porous nature of this material. With human skin in vivo, moisture contained within the body can be secreted via the sweat glands through the natural perspiration process. However, the SC forms a barrier that is only semi-permeable, allowing only gradual movement of water from within the body through the SC into the atmosphere through transepidermal water loss (TEWL) [50]. The SC thus prevents moisture from being squeezed out directly from underlying tissue. A barrier to internal moisture loss is not present with the synthetic skin material, which has a more porous nature. It can therefore be expected that when the synthetic skin is in the moist condition, saturated with water, an applied pressure on the flat rotating ring promotes expulsion of the moisture contained within the bulk of the material, leading to an increased availability of fluid at the surface. It is reasonable to expect that this will increase boundary lubrication and hence reduce the shear strength at the interface (τ) and thus the adhesion friction F_{adh} according to equation (1). If sufficient moisture migrates to the surface, the fluid availability may become sufficient to enable elasto-hydrodynamic lubrication in the boundary to mixed regime, lowering the friction even further.

The results for the sliding-time dependency of the friction for the synthetic skin in moist conditions, Figure7d, lend further support to the hypothesis of pressure-induced migration of fluid to the surface of this material. For a given porosity, the transport of fluid through the material would be time-dependent and moisture would be squeezed out more rapidly at higher surface pressures. This would explain the observation that a decrease in friction with increasing sliding time was only observed at the highest applied surface pressure on the synthetic skin in moist conditions.

Effect of sliding time

Dry human skin in vivo showed a negligible or small increase in friction with sliding time, whereas there was a much more prominent increase in the friction of moist human skin over the test period, see Figures 5 and 8. An increase in friction with sliding time for human skin when using a rotating ring friction torque apparatus, where the skin surface under test is occluded by the test device, has been noted in previously published work [6]. This may be due to a combination of several effects.

Firstly, occlusion of the skin surface by the friction measurement device may have led to increased hydration of the SC by reducing the evaporation rate of secreted perspiration. The higher temperature and humidity of the test environment used for the moist skin can be expected to increase this effect compared to dry conditions because the profuseness of sweating is greater in more humid conditions. As mentioned earlier, increased hydration results in increased adhesion friction according to equation (1). A similar hypothesis has been postulated by Dzidek et al. In experiments with a human finger sliding in a reciprocating manner against smooth glass and polypropylene, the authors found that the coefficient of friction increased by up to an order of magnitude within an occlusion time of 20 s but that this effect was less for higher roughness of the skin-contacting material [51].

A second factor affecting the change in friction with sliding time could be that some abrasion of

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the SC occurs as a result of sliding of the steel ring on the skin surface. Because of the continuous contact between the ring and the skin surface, the debris from this abrasion, consisting of displaced loose corneocytes and other organic matter, is likely to remain largely within the contact and, over time, accumulate and lead to an increased real area of contact and thus increased adhesion friction. Among others, the work of Adams et al [7] suggests that the interfacial shear strength of the SC is reduced by hydration and plasticization. It is therefore reasonable to assume that the SC is more easily abraded and produces more debris in the more hydrated moist state than in the dry state. However, a complicating factor with this hypothesis is that the effect of the reduction in interfacial shear strength on the adhesion friction would compete with the effect of the increase in contact area. For the hypothesis to be true, the latter effect would need to dominate. Further work is clearly needed to establish the roles of these different possible effects in different skin hydration conditions.

Thirdly, because the nominal contact area over which the flat ring was sliding during the test remained constant, local frictional heating of the skin surface is likely and the human skin was indeed reported as “uncomfortably warm” by the test subject in cases where the friction was particularly high. Frictional heating may act in synergy with the skin occlusion effect, promoting further excretion of moisture by sweat glands and increasing the associated plasticisation of SC, leading to a further reduction in the effective elastic modulus.

Regarding the synthetic skin, except for the friction decrease as a function of sliding time discussed earlier for the moist conditions, the synthetic skin was on average less sensitive to sliding time than the human skin, see figures 5, 6 and 8. The reasons for this are unclear. Because perspiration is not possible in the case of synthetic skin, it may indicate that perspiration is the most dominant factor affecting friction increase with sliding time in human skin, or that the surface of the synthetic material is less susceptible to abrasion than human SC and that it does not soften significantly as a result of the frictional heat generated in the test.

Effect of surface pressure

Both the human and synthetic skin showed a decrease in CoF with increasing surface pressure. Assuming the adhesion friction mechanism is dominant, this may be explained on the basis of changes in the real area of contact as a function of surface pressure.

The micro-topography of the skin can be described as a regular two-dimensional wavy surface determined by the primary and secondary lines. The primary and secondary lines form a network like structure of distinct asperities which have a polygon shape [52]. The geometry of the polygon shaped asperities can be idealized as a two-dimensional waviness with wavelength λ and amplitude Δ. A measurement of the skin surface topography of the volar forearm of a healthy 29 year old Caucasian female is shown in Figure 10a, in which the regularity of the micro-topography is evident. The data was collected from a positive replica of the skin and using a Keyence Digital Microscope VHX-2000E. Similar data for the synthetic skin is shown in Figure 10b. Figure 10c shows that the average wavelength and amplitude for the human skin are of the order of 250-450 µm and 20-30 µm respectively, which compare favourably with population-average values reported by Bazin and co-workers of 389 µm 65 µm respectively, for a population of 15 male and 14 female subjects with ages ranging from 7 to 57 years [53]. The average wavelength of the synthetic skin is of the order of 450-750 µm and therefore slightly higher than that measured on the human skin. The average amplitude was of the same order as that measured for human skin, 20-40 µm. This indicates that the synthetic skin reproduces the natural surface

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micro-topography of human skin quite well.

The contact mechanics model for the deformation of a linear elastic two-dimensional wavy surface against a rigid flat surface described by Johnson [54] can be used here to analyse the deformation of the skin surface against the rigid flat surface of the ring probe. For this particular geometry, Johnson shows how the ratio of the real area of contact to the nominal contact area varies with the normal load. For a constant skin hydration state, using our symbols and omitting constant material and geometrical parameters, this relation is:

$$\frac{A_R}{A_p} \propto W^{2/3} \qquad (2)$$

where A_R is the real area of contact, A_p is the nominal contact area and W is the normal load.

The load dependency of the friction is commonly described using the load index, n [7]:

$$CoF = k \cdot W^{(n-1)} \qquad (3)$$

where k is a load-dependent CoF equal to the CoF at unit normal load (W=1), and n is the load index.

Substituting equation (2) in equation (1), and using CoF = F_{adh} / W and τ = constant, reveals that the coefficient of friction depends on the normal load according to:

$$CoF \propto W^{-1/3} \qquad (4)$$

The load index in equation (4) corresponds to n = 2/3.

The load index (n-1) can be determined by plotting log (μ) against log (W) and fitting a power law line through the points. A value of n=1 (i.e. n-1 = 0) would indicate no load dependency of the CoF, i.e. Coulomb friction behaviour. A value of n<1 indicates that the CoF decreases with increasing load, a smaller value of n indicating a greater dependency.

At three, the number of load levels used in the current study is clearly too few to allow statistically sound conclusions to be made. Nevertheless, the analysis shown in Figure 11 shows an average value n = 0.63 for the dry human skin condition, which is very close to the theoretical value of n = 0.66. For this analysis the average CoF measured at the start of the test was used in order to avoid confusion from any possible time-dependent friction changes as a result of frictional heating or otherwise. A similar analysis was not possible for the moist condition because of the rapid increase in friction from first application of the friction device on the moist human skin.

For the same conditions, figure 11 shows that the synthetic skin gave a much lower load index of n = 0.2, indicating a stronger load dependency than with human skin. In this case, this may be attributable to an additional effect of the previously-mentioned phenomenon of moisture contained within the bulk of the material being expelled to the surface under the influence of pressure. The data from five independent drying curves shown in figure 12 indicates that even for a Corneometer® measurement of 20-30 AU, corresponding to the nominally dry condition, the synthetic skin can contain up to 70-80 weight percent water. Thus, although the surface of the material showed a hydration value comparable to that of the human skin in the dry condition (see Table 1), because drying occurs unevenly the material can still contain moisture deeper within the tissue that is not measured by the Corneometer®, which has an effective measurement depth of only 10-20 µm [25]. If the pressure applied to the rotating ring is increased, moisture contained within the bulk of the synthetic skin may be forced to the surface where it can act as a lubricant. Such a phenomenon can be expected to increase the load dependency of the friction.

SUMMARY AND CONCLUSIONS

The friction behaviour of the synthetic skin tested here was notably different to that of human skin in vivo and, for the purpose of using this material as a test-bed in the development of devices that are in sliding contact with the skin, it does not provide an acceptable model of the tribological behaviour of human skin in dry and moist/wet conditions.

The main differences in friction behaviour observed between human skin in vivo and the synthetic skin were:

- The overall magnitude of the CoF of the synthetic skin was much higher than with human skin in vivo. It is likely that this is due to a lower effective elastic modulus of the synthetic skin compared to human stratum corneum, leading to a greater real area of contact with the flat sliding ring and thus higher adhesion friction.
- A change in hydration conditions from dry to moist led to a large increase in the CoF of human skin in vivo but to a large decrease with the synthetic skin. This may be attributed to the porous nature of the synthetic material, leading to an increased availability of fluid at the surface that can be expected to reduce the interfacial shear strength at the interface and hence the adhesion friction.
- Except at the highest applied surface pressure in moist conditions, the friction of the synthetic skin was on average less sensitive to sliding time than human skin. At “high” applied surface pressure in moist conditions, the friction of the synthetic skin decreased with increasing sliding time, an effect not observed with human skin. This may again be attributed to the difference in porosity between the synthetic material and the semi-permeable SC barrier present on human skin.
- Increasing surface pressure or load led to a notable decrease in CoF with both human skin

and the synthetic skin but the effect was greater with the latter material. The load dependency of the CoF for human skin can be explained on the basis of the deformation of the skin surface micro-topography using a simple contact mechanics model. It is hypothesised that the greater load dependency of the CoF of the synthetic skin in dry conditions is again the result of improved boundary lubrication due to pressure-induced migration of moisture from within the moist bulk of the material to the dry surface, whereby the effect increases with increasing pressure.

With human skin in vivo:

- Human skin tended to show an increase in CoF with increasing sliding time, especially in moist conditions. This may be attributed to two effects: a time-dependent increase in SC hydration as a result of occlusion of the skin surface by the friction test device, which may be compounded by frictional heating, and a time-dependent accumulation of abrasion debris between the sliding ring and the skin surface, increasing the effective real area of contact.

Limitations of the current study and future research

Further study is required in order to fully investigate the effects of skin surface occlusion, frictional heating and SC abrasion on the development of skin friction as a function of sliding time. The effects of occlusion and frictional heating can be investigated by carrying out friction experiments with the rotating-ring test device with and without ring rotation, with close monitoring of the skin temperature during and after testing. To investigate the effects of SC abrasion, the debris produced as a result of sliding the ring on the skin surface should be analysed, whether this debris transfers and becomes attached to the sliding ring, and how this

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affects the effective roughness and contact area of the ring. Any resulting changes in the contact area need to be considered together with hydration-related changes in the interfacial shear strength of the SC, because of the opposing effects of these factors on the adhesion friction. Regarding a synthetic skin model, in order to be useful as a tribological test-bed for use in the development of devices that slide against the skin in different states of hydration, a model incorporating at least two layers is required. The elastic modulus of the human stratum corneum or the complete epidermis and the ability of the stratum corneum to become hydrated, whereby the elastic modulus is reduced and the real area of contact is increased, needs to be simulated by the uppermost layer of the synthetic material. A hydrophilic material may be suitable here, one that absorbs and releases moisture in a similar way to the SC of human skin so as to mimic the effect of plasticization on the real surface area of contact, and that has an intrinsic interfacial shear strength similar to that of human skin. The underlying layer should have a much lower elastic modulus to simulate the effective elastic modulus of the human dermis and hypodermis and the effective deformation behaviour of the composite human skin structure. This layer should not change its properties through absorption of moisture from the environment and no migration of fluid to the surface of the top layer, where it could affect the boundary lubrication behaviour, should occur. Such behaviour could be achieved, for example with silicone- or polyurethane-based elastomers or a sealed hydrogel in which the water is trapped so that it cannot be squeezed out when pressure is applied.

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Ethical approval

All procedures performed in studies involving human participants were in accordance with the ethical standards of the institutional and/or national research committee and with the 1964 Helsinki declaration and its later amendments or comparable ethical standards.

Informed consent

Informed consent was obtained from all individual participants included in the study.

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Table 1: Skin characteristics and test conditions.

		Dry skin condition		Moist skin condition	
Type of skin	**Characteristics**	**Skin hydration [AU]**	**Surface temperature [°C]**	**Skin hydration [AU]**	**Surface temperature [°C]**
Human skin	Left inner hairless forearm of one healthy 24 yr. old Caucasian female	21-28	25-29	69-80	31-32
Synthetic skin	Commercial SynTissue™ “Adult skin” model from Syndaver™ Labs, SKU: T-SKN-A-0005. 1.2 mm thickness, 2 N puncture force in the manufacturers own standard penetration test, “Caucasian” colour [24].	18-29	22-25	61-78	29-30

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Table 2. Summary of tribological test conditions.

Ring	Material: stainless steel X10CrNiS 18 9. Roughness R_a 0.29µm (SD: 0.03µm). Surface area 62.8mm^2.
Load on ring [N]	0.31, 0.94 and 1.56
*Nominal surface pressure [kPa]	4.9 (“light”), 15.0 (“medium”) and 24.8 (“high”)
Ring rotation speed	2.56 revs/s = 96 mm/s at outer radius of ring
Sliding time	20 – 30 s

*Assuming full contact between the ring and the skin or synthetic skin.

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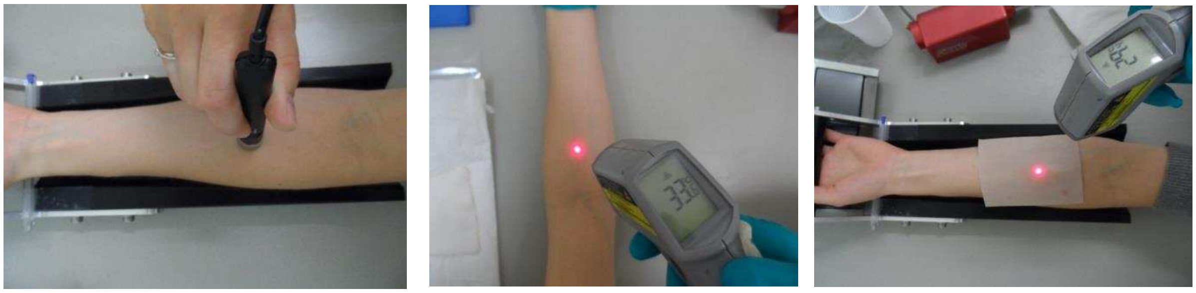


a) b) c)

Figure 1: Measurement of a) skin hydration level on human skin; b) skin surface temperature on human skin; c) surface temperature on synthetic skin.

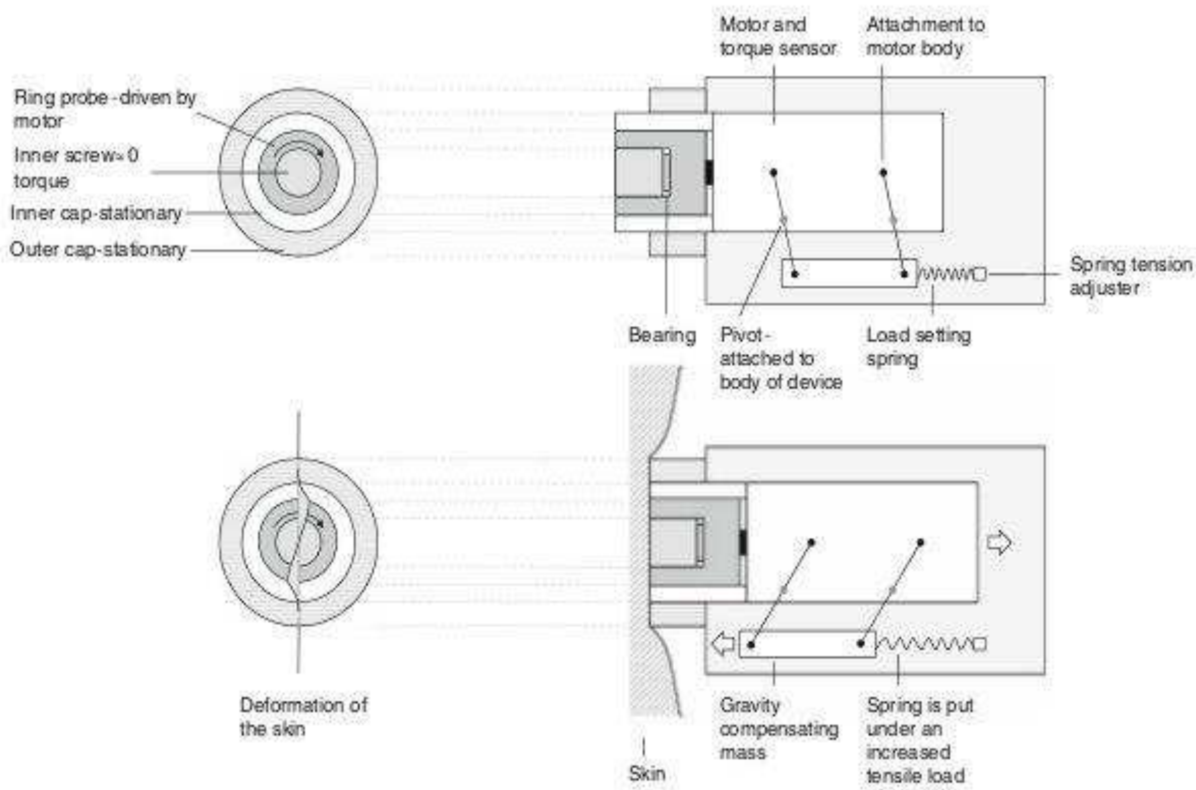


Figure 2: Rotating ring friction measurement device: schematic [6]

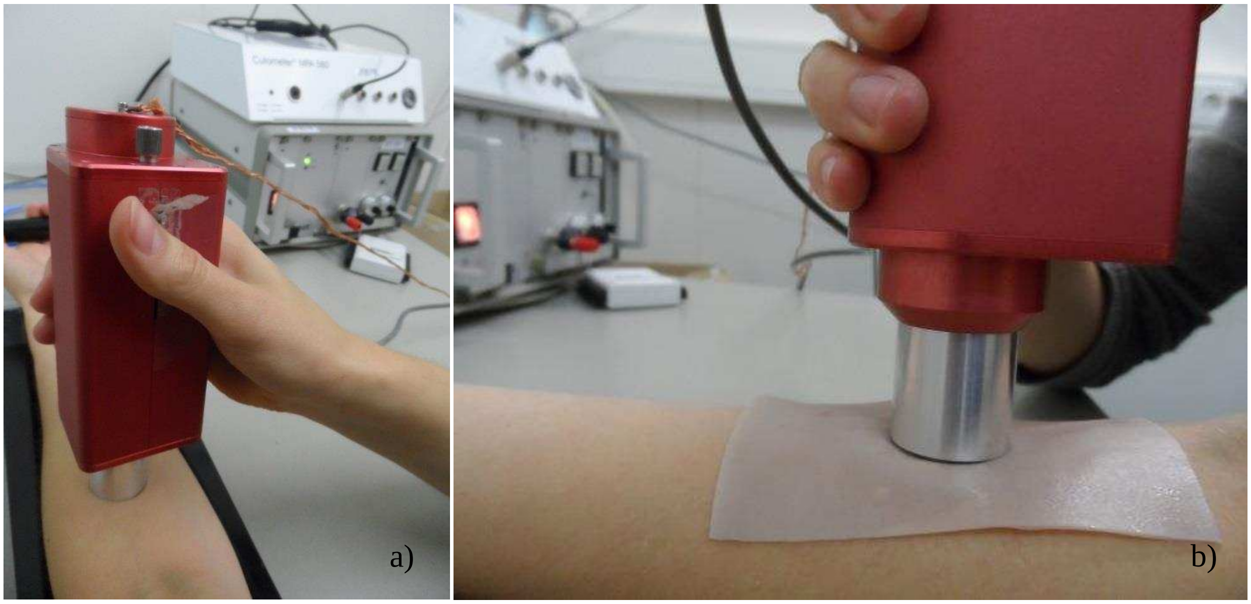
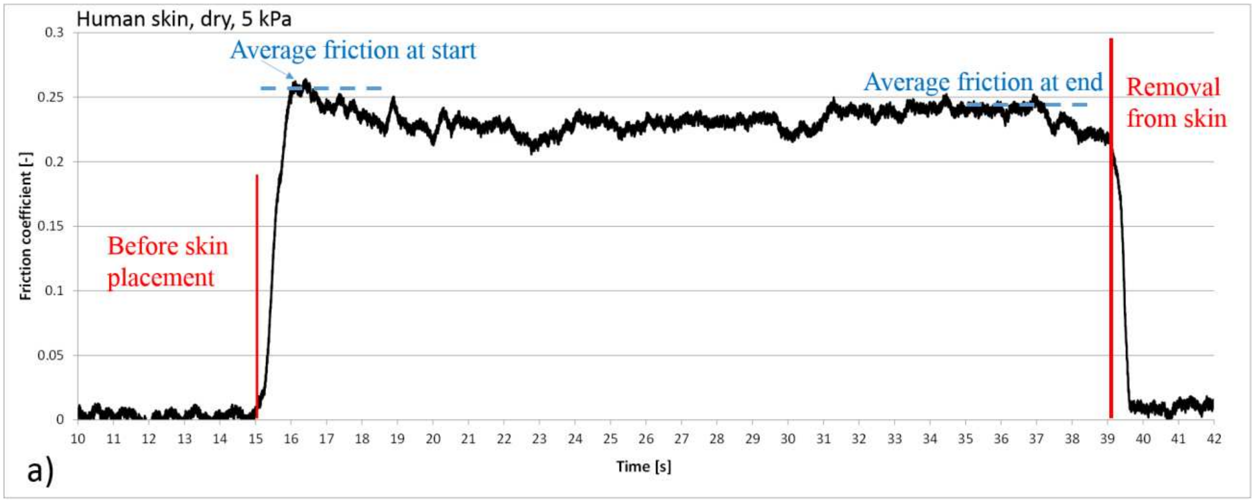
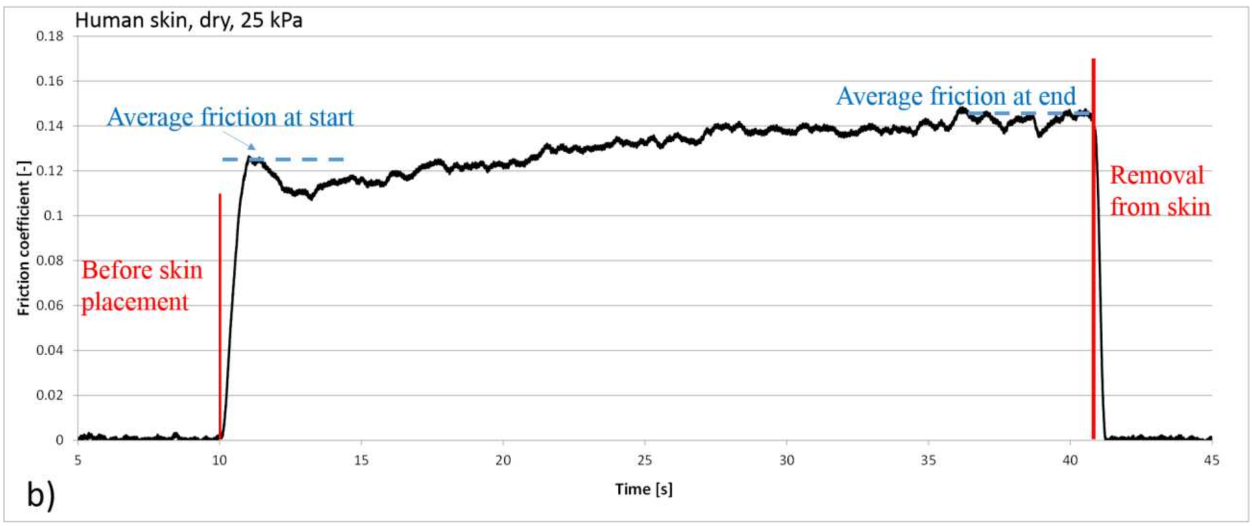


Figure 3: Placement of rotating ring friction device on a) human skin in vivo and b) synthetic skin.





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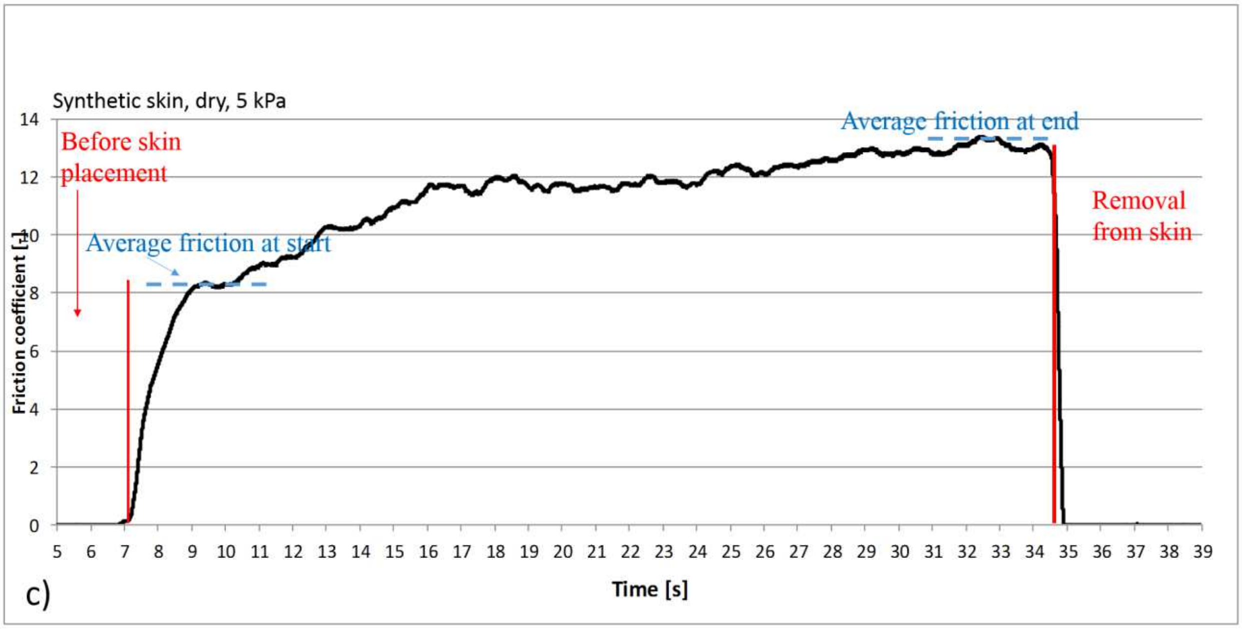

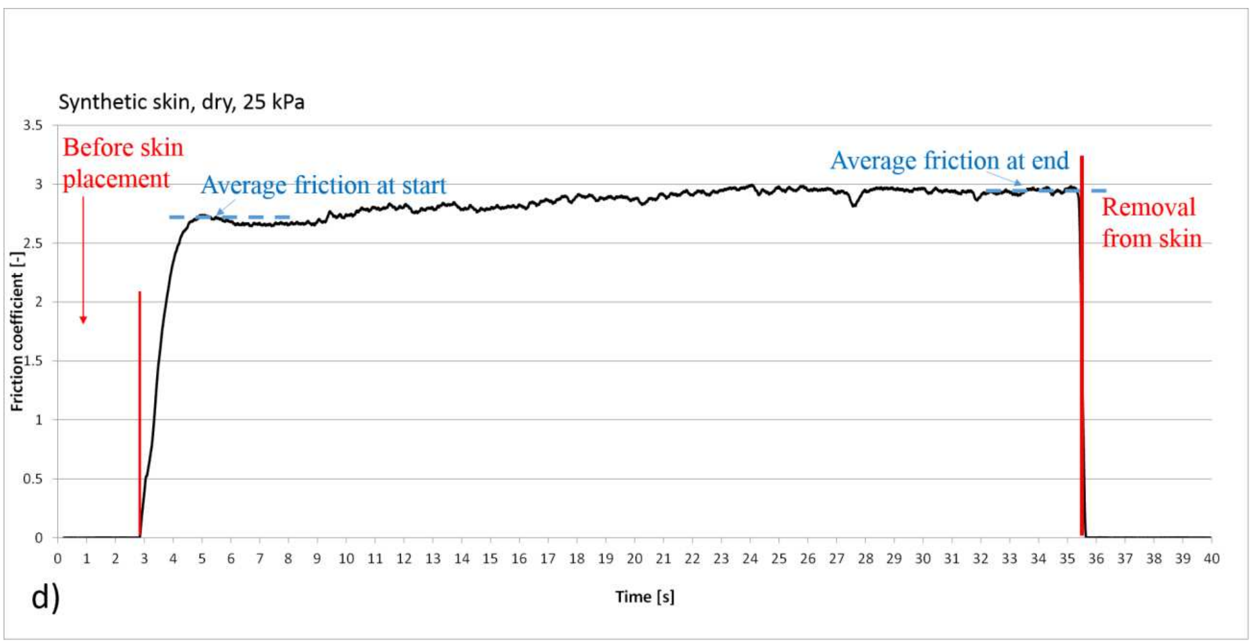

Figure 4 Characteristic friction behaviour in dry conditions of a) Human skin at a surface pressure of 5 kPa; b) Human skin at a surface pressure of 25 kPa; c) Synthetic skin at a surface pressure of 5 kPa; d) Synthetic skin at a surface pressure of 25 kPa.

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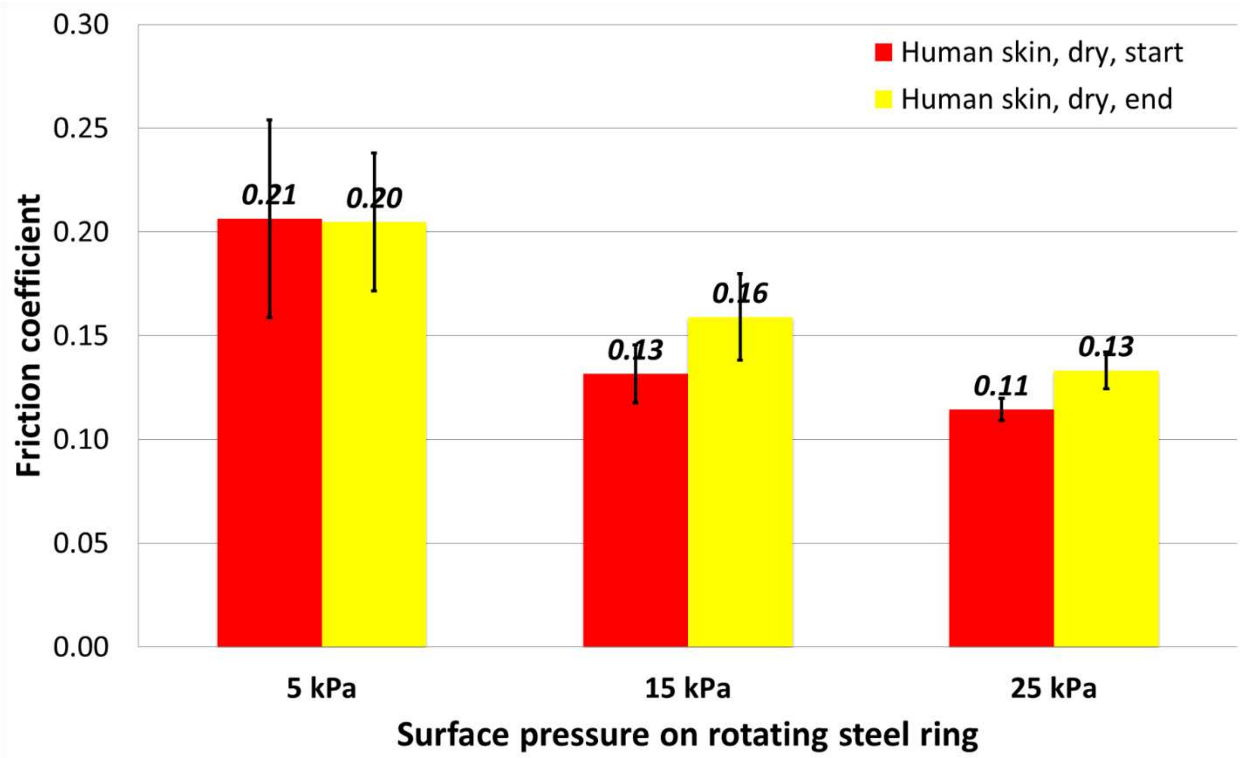


Figure 5: Average friction coefficient for human skin in dry conditions at test start and end (respectively, a period of 0.5 to 1s after first applying the friction device to the skin and the final 0.5 to 1s of testing). The error bars represent one standard deviation.

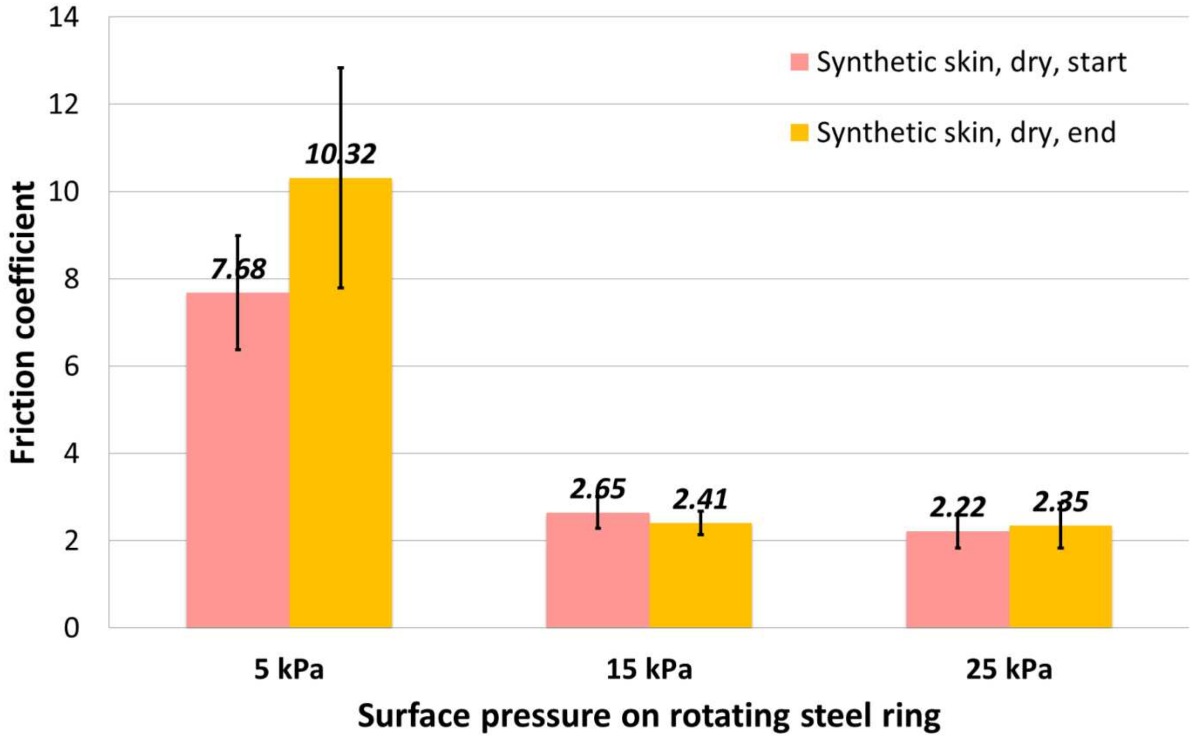
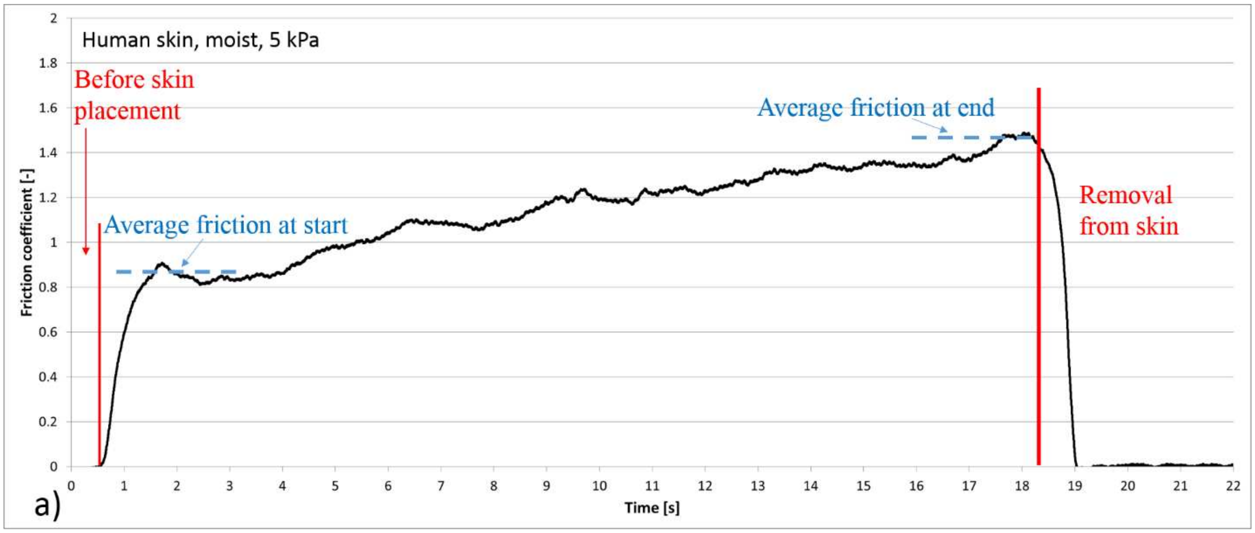
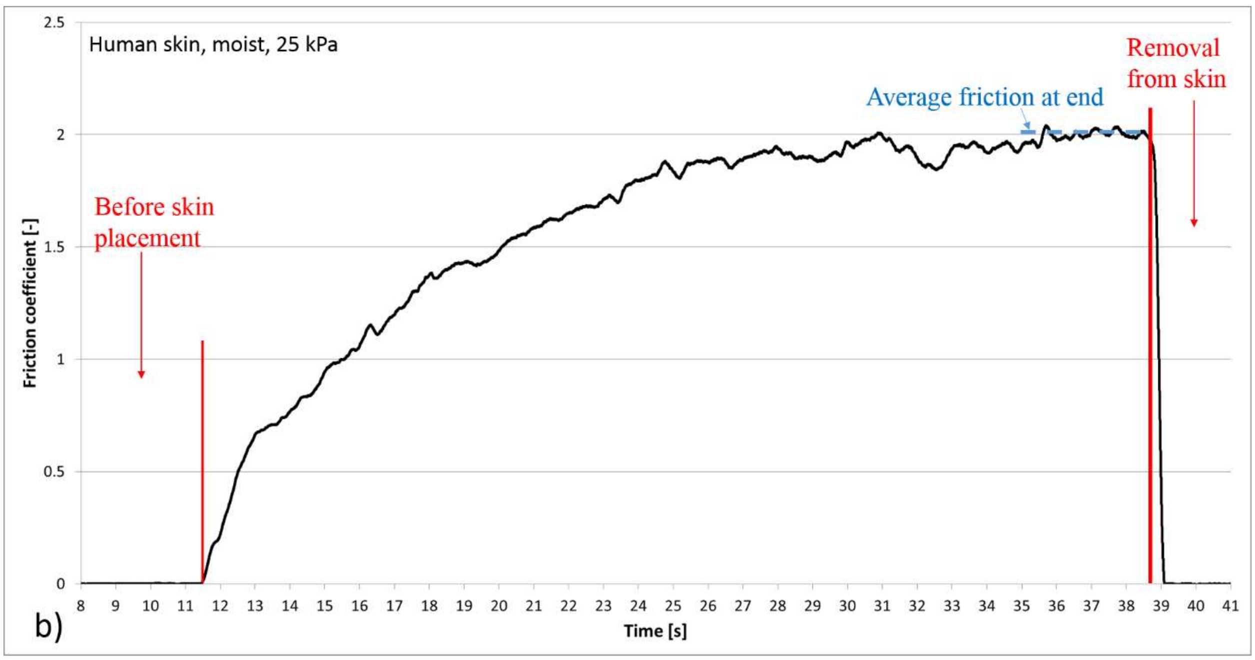


Figure 6: Average friction coefficient for synthetic skin in dry conditions at test start and end (respectively, a period of 0.5 to 1s after first applying the friction device to the skin and the final 0.5 to 1s of testing). The error bars represent one standard deviation.


Human skin, moist, 5 kPa
Before skin placement
Average friction at start
Average friction at end
Removal from skin
Friction coefficient [-]
Time [s]
a)


Human skin, moist, 25 kPa
Removal from skin
Average friction at end
Before skin placement
Friction coefficient [-]
Time [s]
b)

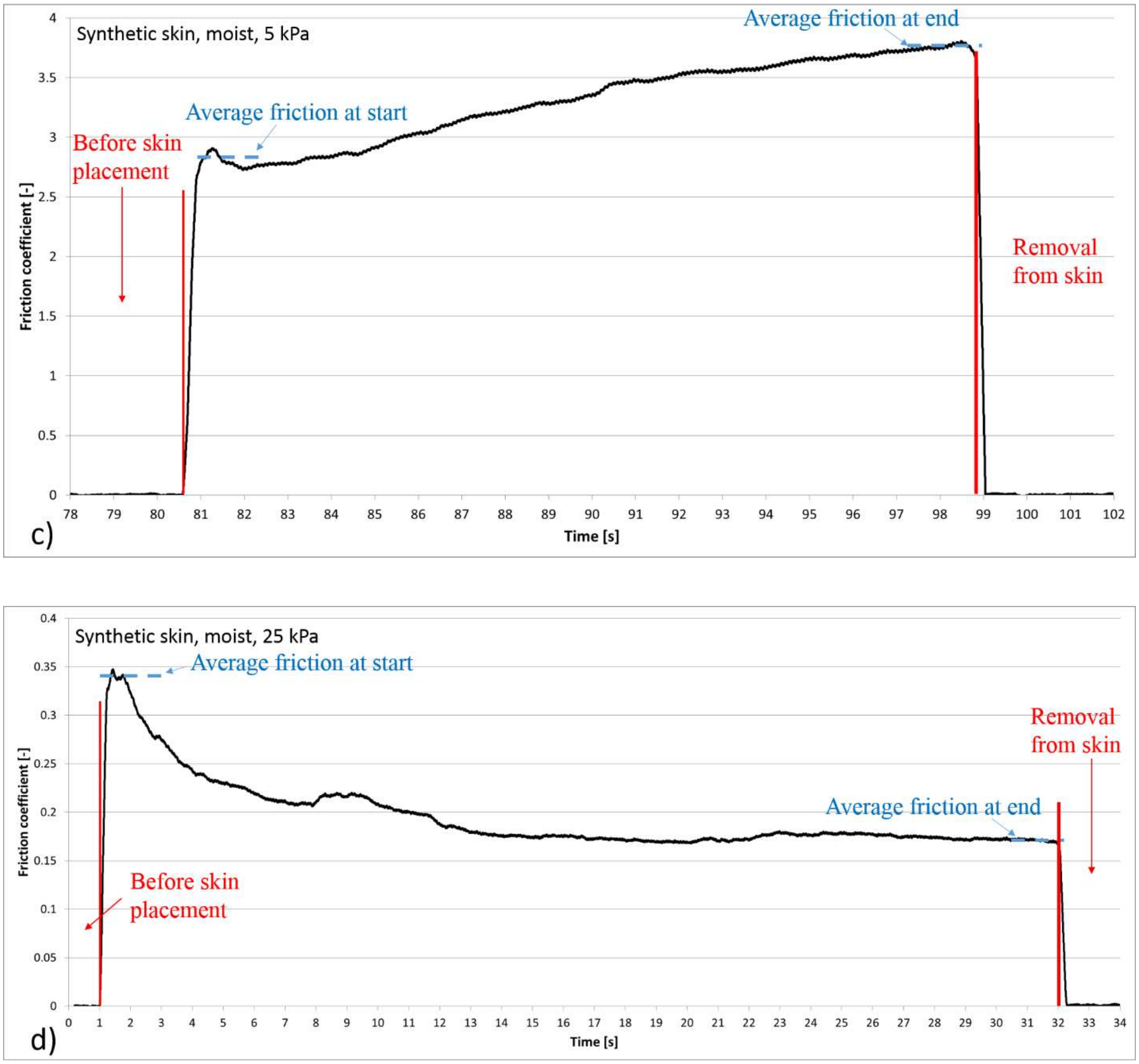


Figure 7. Characteristic friction behaviour in moist conditions of a) Human skin at a surface pressure of 5 kPa; b) Human skin at a surface pressure of 25 kPa; c) Synthetic skin at a surface pressure of 5 kPa; d) Synthetic skin at a surface pressure of 25 kPa. Note the lack of a clear starting point for the friction at the start of the test for the human skin at a surface pressure of 25 kPa.

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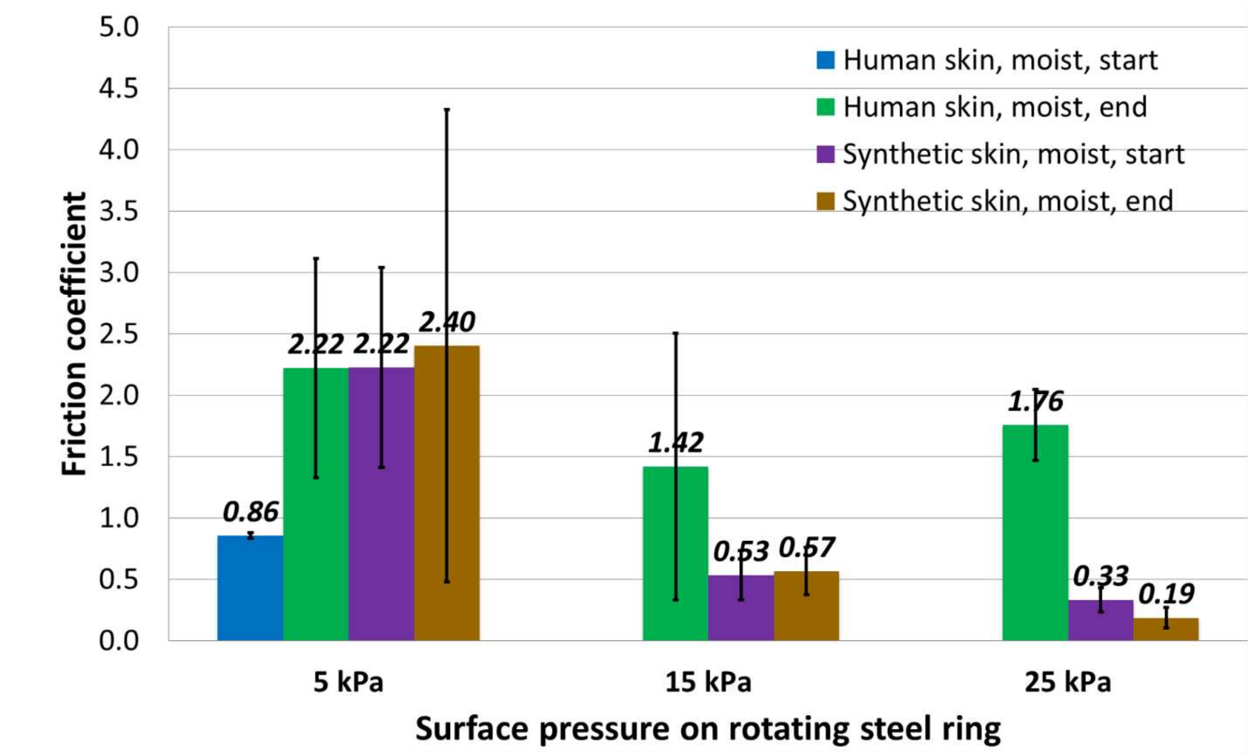

Figure 8. Average friction coefficient for human and synthetic skin in moist conditions at test start and end. The error bars represent one standard deviation. Note that it was not possible to observe a clear starting friction value for human skin at 15 and 25 kPa because the friction began to increase immediately from first skin contact.

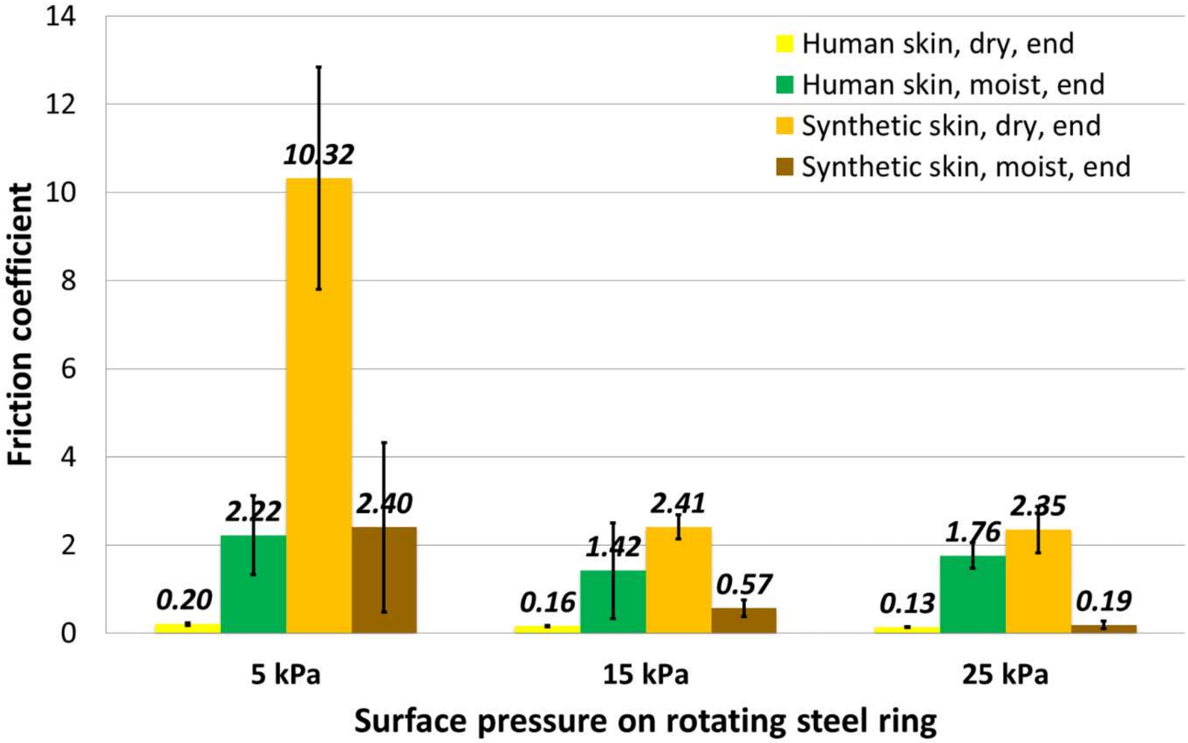

Figure 9. Average friction coefficient for human and synthetic skin in dry and moist conditions at test end (the final 0.5 to 1s of testing). The error bars represent one standard deviation.

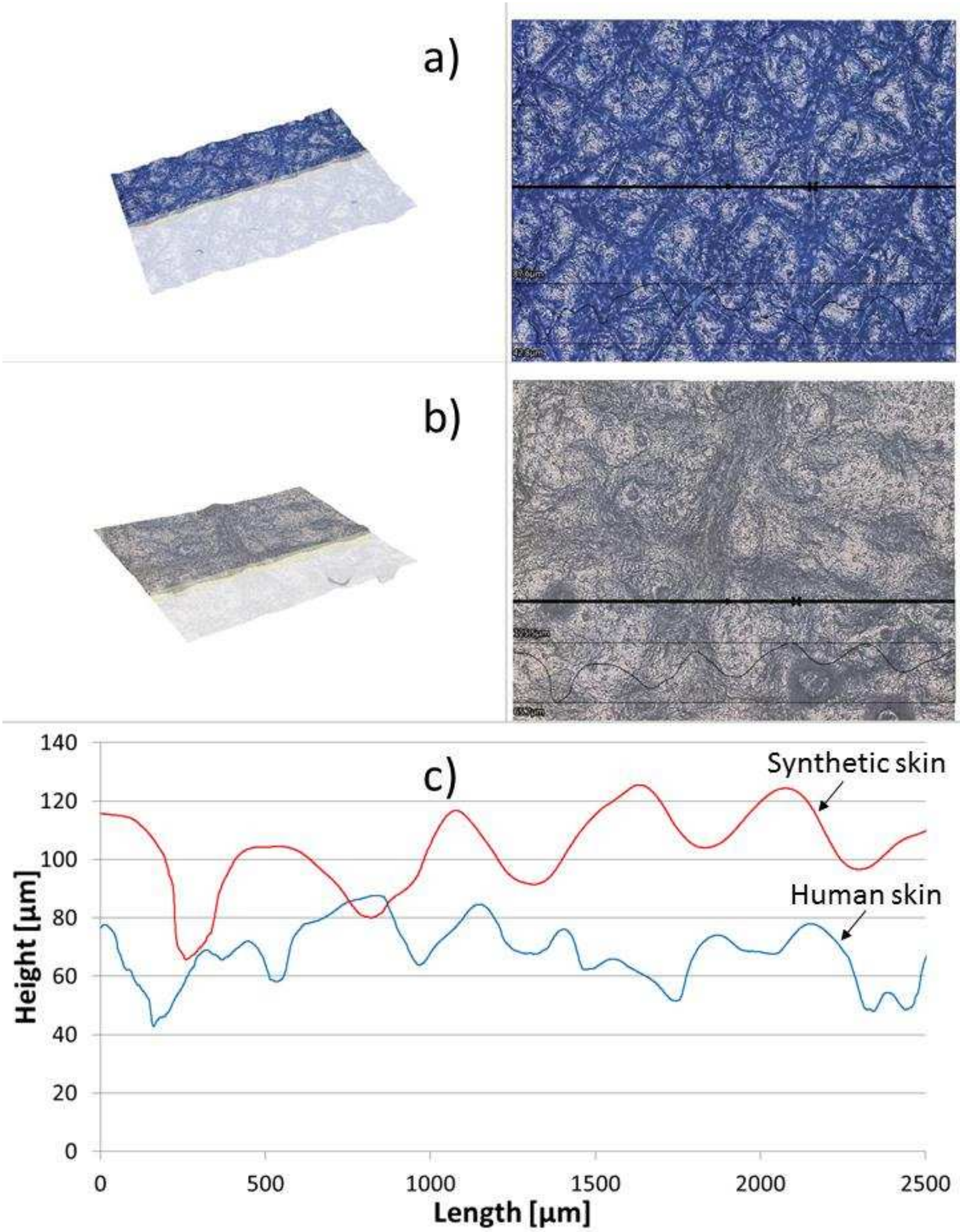

Figure 10: Surface micro-topography of a) the human skin of the volar forearm of a healthy 29 year old Caucasian female and b) the synthetic skin. c) shows data from a line scan across the digital images.

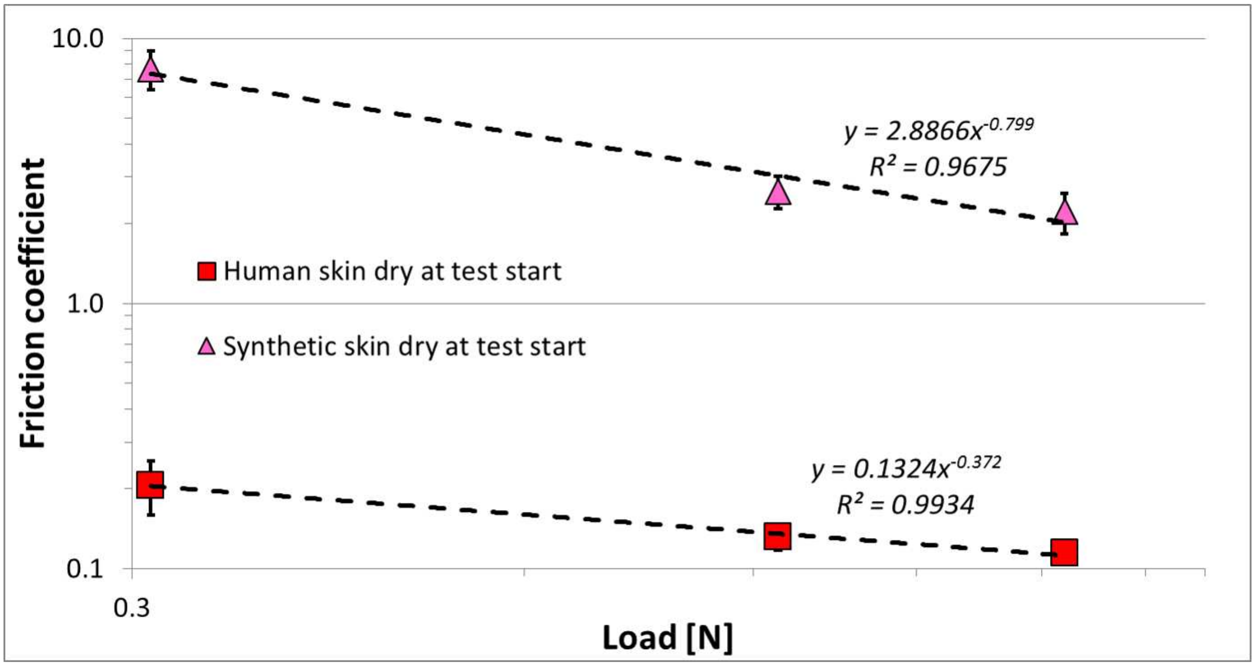

Figure 11. Log-log plot of the average coefficient of friction at the start of the test as a function of applied load for dry human and synthetic skin conditions. The error bars represent one standard deviation.

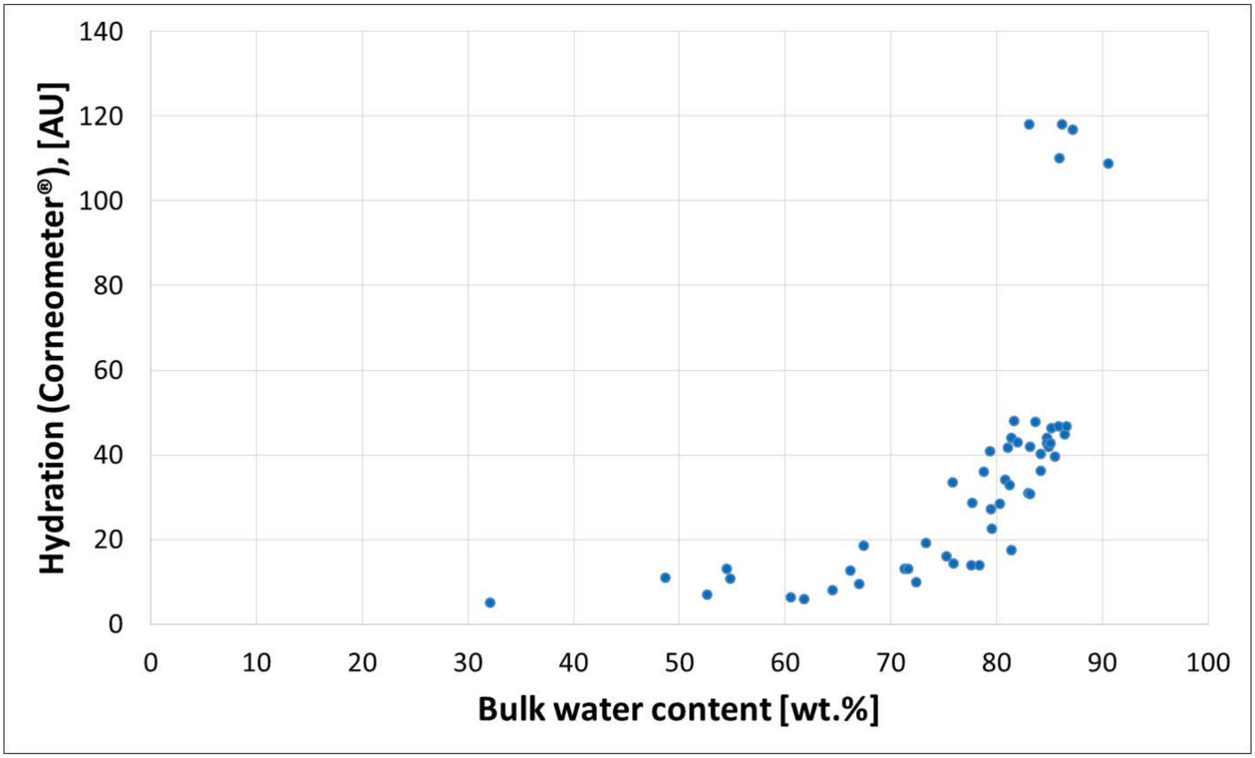

Figure 12. Data from five independent measurements of the surface hydration of synthetic skin as a function of actual bulk water content (measured on the basis of weight loss during drying).