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ORIGINAL ARTICLE

Effects of humidity on skin friction against medical textiles as related to prevention of pressure injuries

Danit Schwartz | Yana Katsman Magen | Ayelet Levy | Amit Gefen

Department of Biomedical Engineering, Faculty of Engineering, Tel Aviv University, Tel Aviv, Israel

Correspondence
Prof. A Gefen, Department of Biomedical Engineering, Faculty of Engineering, Tel Aviv University, Tel Aviv 69978, Israel.
Email: gefen@eng.tau.ac.il

Sustained pressure, shear forces, and friction, as well as elevated humidity/moisture, are decisive physical factors in the development of pressure injuries (PIs). To date, further research is needed in order to understand the influence of humidity and moisture on the coefficient of friction (COF) of skin against different types of medical textiles. The aim of this work was to investigate the effects of moisture caused by sweat, urine, or saline on the resulting COF of skin against different textiles used in the medical setting in the context of PI prevention. For that purpose, we performed physical measurements of static COFs of porcine skin followed by finite element (FE) computational modelling in order to illustrate the effect of increased COF at the skin on the resulting strains and stresses deep within the soft tissues of the buttocks. The COF of dry skin obtained for the 3 textiles varied between 0.59 (adult diaper) and 0.91 (polyurethane dressing). In addition, the COF increased with the added moisture in all of the tested cases. The results of the FE simulations further showed that increased COF results in elevated strain energy density and shear strain values in the skin and deeper tissues and, hence, in an increased risk for PI development. We conclude that moisture may accelerate PI formation by increasing the COF between the skin and the medical textile, regardless of the type of the liquid that is present. Hence, reduction of the wetness/moisture between the skin and fabrics in patients at a high risk of developing PIs is a key measure in PI prevention.

KEYWORDS

computational modelling, finite element analysis, pressure injuries, skin friction

1 | INTRODUCTION

According to current international guidelines, a pressure injury (PI), also known as a pressure ulcer, is an injury localised to the skin and/or underlying tissues caused by pressure or pressure in combination with shear.[1] PIs may initiate either superficially or in deep tissues and most commonly develop over bony prominences or when the skin is in contact with rigid medical devices.[2,3] The elderly, neuromuscular patients, trauma patients, and patients in lengthy operations have a high risk of developing PIs; however, injuries may also be inflicted by misuse of medical devices.[4] Prolonged pressure, shear forces, and friction, as well as elevated humidity/moisture, are decisive physical factors in the development of PIs. Therefore, the reduction of friction and moisture (which are inter-dependent) at the skin-textile interface is a central goal in the prevention of skin and deeper injuries.[5]

It is generally acknowledged that the frictional properties of human skin are determined by the surface properties of the skin itself (roughness, hydration state etc.) and the properties of the contacting materials (stiff, soft, fibrous etc.), as well as possible intermediate layers such as topically applied substances, for example, creams and moisturisers, or sweat and sebum, which are naturally excreted from skin into the tribo-interface.[6–8] Fabric-skin friction typically does not obey Amontons' law, where friction force is directly proportional to normal force and independent of the apparent contact area. In the case of fabric-skin contact, the friction forces that develop at the interface strongly depend on the level of wetness and are a combination of the forces required to break the adhesive bonds between the 2 surfaces and the forces related to the deformations of the bodies in contact.[9–11] Adhesion is currently considered the main contributor to the friction of human skin, whereas deformation mechanisms are assumed to play a less dominant role. The coefficient of friction (COF) between human skin and fabrics is influenced by the nature of the textile materials, the contact pressure, mechanical skin properties, type of relative motion, the ambient humidity, and the skin moisture content.[12–15] Static COF is measured at the point where an object starts sliding against the skin, whereas dynamic COF is measured while the same object is already moving. Static frictional forces between 2 surfaces will increase to prevent any relative motion up until a certain point where motion initiates. It is that threshold of motion which is characterised by the coefficient of static friction.

Several physical mechanisms have been studied as contributors to the increased COF of skin in humid or wet environments: swelling and softening of the stratum corneum (SC), viscous shearing of liquid bridges formed between the skin and the interacting surface, the work of adhesion because of absorbed moisture, and, finally, the formation of a glue-like layer because of the solution of skin lipids and proteins in a thin layer of absorbed water or sweat.[7] In a pilot study from 2011, Tomlinson aimed at quantifying, for the first time, the relative contributions of water absorption, capillary adhesion, and viscous shearing effects on skin friction in moist conditions.[16] Using finger friction measurements on a polyvinylchloride plate, they concluded that water absorption is the main mechanism responsible for the increase in friction, followed by capillary adhesion, although it was not conclusively proven that the latter contributed significantly. If skin is saturated with water, and excess water accumulates at the interface, capillary bridges between the skin and the interacting surface might be relevant to a certain degree, but with further increasing amounts of water, lubrication phenomena will become more pronounced.[17] Hydrodynamic lubrication (EHL) is characterised by the complete separation of the sliding surfaces by a liquid film. Under such conditions, the adhesion component of friction is replaced by viscous friction. Depending on contact conditions as well as fluid film thickness in relation to the surface roughness of the skin and the contacting material, mixed lubrication or boundary lubrication can also take place. The former lubrication regime is characterised by the coexistence of dry and wet contact zones, the latter by molecular surface films influencing the frictional behaviour. However, the contributions of EHL alone were found to be too small to fully explain the complex frictional behaviour of wet skin.[11,16–18]

The pig is a well-accepted skin model as its integument is morphologically and functionally similar to human skin.[19] As with all animal models, similarities and differences between pigs and humans coexist. Macroscopically, much like humans, the pig is a relatively hairless animal with a fixed skin that is tightly attached to the subcutaneous tissues. The cutaneous blood supply and sequence of events in wound healing are also similar to those in humans.[20] Swine and humans have similar body surface areas, which also makes them more compatible than smaller animals, such as rodents. However, porcine skin is overall thicker than human skin, especially at the dorsal surface of the neck and back of sexually mature animals. Additional similarities between the skin of pigs and humans are a relatively thick epidermis, epidermal turnover kinetics, lipid composition, carbohydrate biochemistry, enzyme histochemistry, lipid biophysical properties, and arrangement of dermal collagen and elastic fibres.[20] All of the above facilitate the use of porcine skin for measurement purposes, in the context of PI research, and allow us to use the results as indicators of expected human skin frictional behaviour.[19,20]

Key Messages

- moisture may accelerate pressure injury formation by increasing the coefficient of friction between the skin and the medical textile, regardless of the type of the liquid that is present at the skin-fabric interface
- in this study, we performed physical measurements of static coefficients of friction between porcine skin and different types of medical textiles, in various moisture conditions, followed by finite elements computational simulations to illustrate the effect of increased COF on deep tissue loads
- the COF of dry skin obtained for the 3 medical textiles that were tested varied between 0.59 (adult diaper) and 0.91 (polyurethane dressing), and increased with the added moisture in all of the tested cases
- the results of the FE simulations further showed that increased COF results in elevated strain energy density and shear strain values in the skin and deeper tissues, and hence, increased risk for PI development

Psychrometry is the measurement of the moisture content of air. Atmospheric air is a mixture consisting of dry air and water vapour in varying relative amounts. A 2-bulb psychrometer is a measurement device that includes both dry-bulb and wet-bulb thermometers, where the dry-bulb temperature refers to the ambient air temperature, and the wet-bulb temperature is a thermodynamic property of a mixture of air and water vapour. The wet-bulb temperature is always between the dry-bulb temperature and the dew point (the temperature at which air becomes completely saturated; above this temperature, the moisture stays in the air). Knowing both the dry-bulb and wet-bulb temperatures, one can determine the relative humidity (RH) using a psychrometric table.[21,22]

Finite element (FE) computational modelling is a powerful tool in PI research. This type of modelling allows the quantifying of skin as well as internal tissue deformations, strains, and stresses in weight-bearing body parts, for example, the heels and buttocks.[23–28] In the current study, the FE method was used in order to illustrate the influence of a high versus low COF on the distributions of strains and stresses in deep soft tissues, rather than on just the skin surface.

To date, the influence of humid environment because of sweat, urine, and wound exudate/drainage (which will be modelled here using saline) on skin COF is not sufficiently understood and requires additional research, with an emphasis on the effect of different types of textiles used in medical settings. Hence, the aim of this work was to investigate the effects of different humidity sources on the COF between skin and different medical textiles in the context of PI prevention.

2 | MATERIALS AND METHODS

2.1 | Porcine skin

In this study, fresh porcine skin was used as a model for human skin. Samples measuring 9 × 18 cm were taken from the dorsal area of sexually mature 4- to 6-month-old Yucatan pigs, used for a different (unrelated) acute study (piggyback protocol). In order to measure frictional and mechanical properties that are as close as possible to those of living skin, the experiments described below were performed on manually shaved samples, immediately after euthanasia, thus avoiding freezing and thawing as well as chemical preservation of the samples.

2.2 | Medical textiles

To ensure clinical relevance, different commercially available medical textiles were selected for testing. The investigated fabrics included a standard hospital bed sheet consisting of 100% cotton, with no chemical finishing or coloration; the inner layer of a standard adult diaper (a cotton-like synthetic fabric); and a standard polyurethane single-layer foam dressing (4 mm thickness), which is commercially available and is prescribed for treating existing wounds as well as for prophylaxis.[29,30]

2.3 | Compression testing

In order to verify that the porcine skin specimens were indeed representative of adult human skin, a uniaxial compression test was performed, using an electromechanical testing system (INSTRON Co. model 5544; High Wycombe, UK), at a quasi-static strain rate of 2 mm/min. Five to 7 layers of cylindrical specimens (40 mm in diameter) were stacked between the top and bottom compression plates of the testing machine (Figure 1A). The forces and displacements were recorded, and engineering strains and stresses were calculated to obtain the effective stiffness (elastic modulus E) of the porcine skin from the initial slope of the stress–strain curve.[31–35]

2.4 | Humidity control

To ensure uniform RH conditions across measurement, a 2-bulb psychrometer device was built. The psychrometer consisted of 2 thermometers with an accuracy of $\pm 0.1^{\circ}$C; the first was kept dry, while the second was wrapped with fabric and dipped periodically in distilled water throughout the experiments (Figure 1B). The psychrometer was placed inside a biological fume hood, and the ventilators were set to medium speed in order to create a humidity-controlled environment during the experiments. The RH was calculated using a psychrometric table[21,22] and kept constant at 58% during all measurements. The ambient air temperature was also kept constant at 26 °C.

2.5 | Experimental setup

A tilting-table electric tribometer, which was developed in-house, was used to investigate the frictional behaviour of the aforementioned porcine skin against different medical textiles under the moisture conditions specified above. The tribometer was composed of 2 main parts (Figure 1C): a tilting plate and a circular steel weight. The tilting plate was covered with porcine skin and controlled by an electric motor via a cord. The circular steel weight was wrapped with the textile sample and placed on top of the porcine skin. The edge of the tilting plate was then slowly lowered to increase the angle of inclination θ (Figure 1C). When the threshold of motion was achieved, and the weight started sliding down the tilting plate, the motor automatically stopped, and θ was recorded using a clinometer application installed on a smartphone. The clinometer application was calibrated prior to acquiring every set of measurements. The static COF was then calculated for each measurement as $\tan(\theta)$.

Three circular textiles samples (50 mm in diameter) from each type of the investigated textile were used for the COF measurements. On each textile sample, we performed a set of 6 measurements, which provided a total of 54 measurements. The porcine skin samples were replaced after each set of measurements.

After completing the COF measurements for the reference (referred to as "dry") skin conditions, we investigated the effects of added moisture because of sweat, urine, or saline, at the interface between the skin and the textiles, on the resulting COF. Textile samples were saturated with 8 mg/cm^2 of human sweat, human urine, or saline, which are equivalent to the mean transepidermal water loss in 24 hours.[36] The urine and sweat samples were collected from adult volunteers using sterile beakers and were maintained under cooling conditions. Sweat was collected during gym practice from professional athletes. The COF under

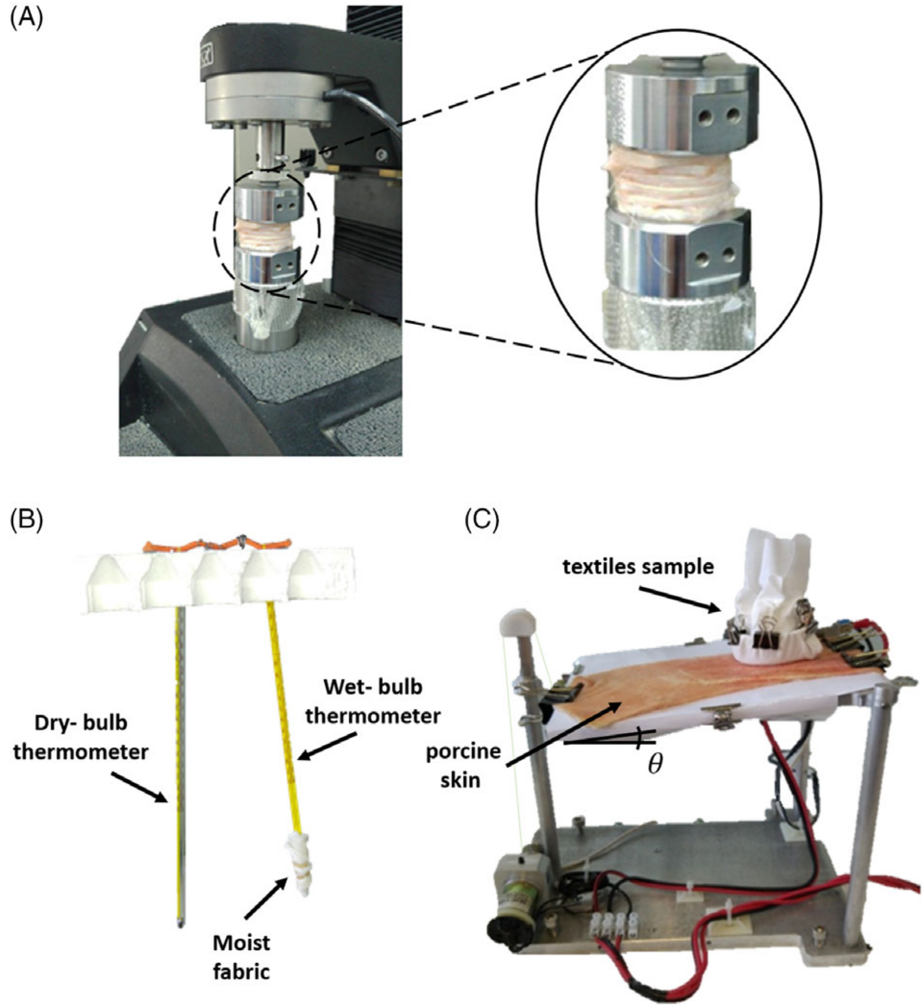

FIGURE 1 Mechanical measurements of the porcine skin. A, The uniaxial compression testing setup. B, The 2-bulb psychrometer used for relative humidity (RH) measurements. C, The tilting-table tribometer used for friction measurements of the porcine skin against the different types of medical textiles

moist conditions was then measured using the same tribometer and experimental technique as described above.

2.6 | Statistical analysis

Statistical analysis of the COF measurements was performed using the online software tool, vasserstats.net. We performed a 2-way ANOVA test to identify significant factors influencing the COF. The 2 factors that have been tested are fabric type and skin condition (dry and wet because of saline, sweat, or urine). Then, we performed a 1-way ANOVA to distinguish between statistically significant factors within groups. For all analyses, statistical significance was set at $P < .05$. Results are expressed as means ± 1 SD from the mean.

2.7 | Computational simulations

In order to illustrate the effect of variations in the COF on the resulting internal strain and stress distributions in the soft tissues of the buttocks, and hence on the risk for developing PIs and especially deep tissue injuries (DTIs), we developed 2 FE computational simulations. We used a graphical representation of the pelvis (Figure 2A) and the ScanIP module of Simpleware[37] to segment the pelvic bones, muscles, fat, and skin tissues and then to define a uniform 3-mm thickness to the entire slice model (Figure 2B). A standard flat foam cushion was added under the buttocks at the pre-processing stage in the PreView module of FEBio.

Constitutive laws and mechanical properties of all the tissues included in the buttocks model were adopted from the literature (Table 1). Specifically, the pelvic bones were assumed to be linear-elastic isotropic material with an elastic modulus of 7 GPa and a Poisson's ratio of 0.3[35] (Table 1). Skin, fat, and muscle tissues were assumed to be nearly incompressible (Poisson's ratio of 0.49) non-linear isotropic materials, with their large deformation behaviour described using a Mooney-Rivlin material model with the following strain energy density (SED) function W:

$$W = \frac{G_{\text{ins}}}{2}\left(\lambda_1{}^2 + \lambda_2{}^2 + \lambda_3{}^2 - 3\right) + \frac{1}{2}K(\ln J)^2 \qquad (1)$$

where G_{ins} is the instantaneous shear modulus, λ_i (i = 1,2,3) are the principal stretch ratios, K is the bulk modulus, and $J = \det(F)$, where F is the deformation gradient tensor (Table 1). We chose material constants based on the values reported by Oomens and colleagues, with appropriate adjustments made by Zeevi et al.[38,39] The support was assumed to be isotropic linear-elastic with a Poisson's ratio

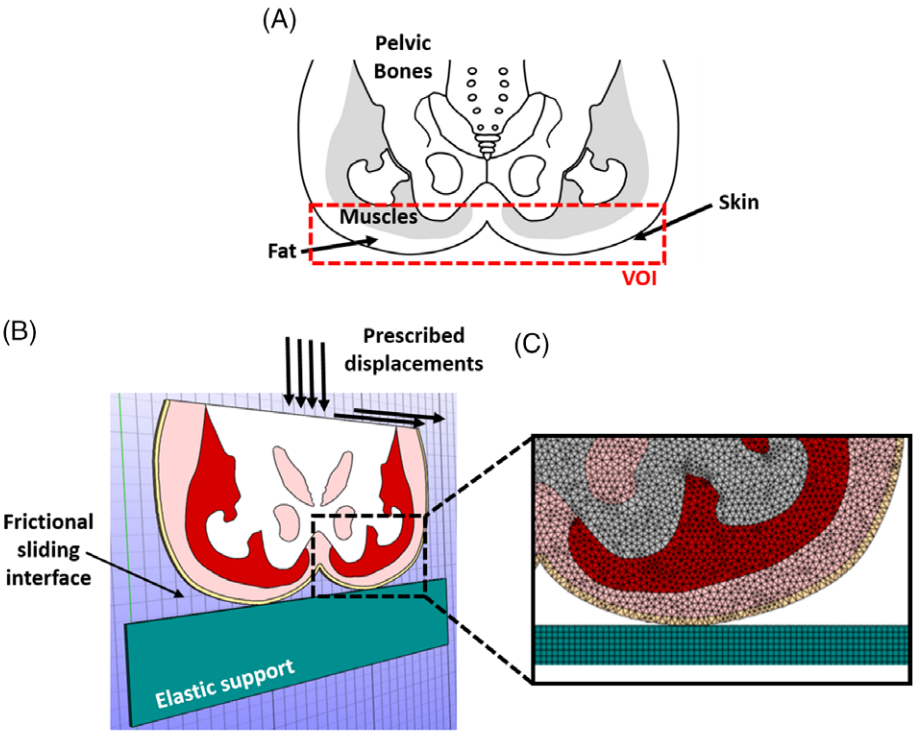

FIGURE 2 Computational model of the seated buttocks. A, A graphical illustration of the buttocks, showing the pelvic bones, muscles, fat, and skin tissues. The volume of interest is marked using a dasher red line. B, The boundary and loading conditions of the thin slice model. C, Zoom-in on the tetrahedral mesh

TABLE 1 Mechanical properties of all the model components: tissues and the support

Model component	Shear modulus (kPa)	Bulk modulus (kPa)	Elastic modulus (kPa)	Poisson's ratio
Skin	8	4000	—	—
Fat	1	500	—	—
Muscle	0.3	150	—	—
Bone	—	—	7×10^6	0.3
Foam support	—		25	0.3

of 0.3 and an elastic modulus of 25 kPa, which is within the stiffness range of standard hospital mattresses.[27]

Boundary conditions were selected to simulate sagging of the buttocks during sitting on a flat elastic cushion in a thin-slice model. In order to represent the shearing forces that may act on the buttocks during performance of daily actions, such as transferring from a bed to a wheelchair, from a wheelchair to a car seat, or when simply changing positions in the chair, we used a combination of compression and shear loading. Specifically, we prescribed simultaneous vertical and horizontal displacements of 10 mm on the top surface of the anatomical model (Figure 2B). The front and back planes of the buttocks and support were fixed for out-of-plane motions to enforce the thin-slice conditions, and the bottom surface of the elastic support was fixed for all translations and rotations. We defined frictional sliding between the support and skin, with the COF set to 0.6 or 1.2 in order to simulate low versus high COF, respectively, which are the minimal and maximal values that were measured in the experiments reported here.

Meshing of the tissues was performed using the ScanIP module of Simpleware,[37] and meshing of the support was performed in the Preview module of FEBio[40] (Figure 2C). Final meshes included 75 364 linear tetrahedral elements describing the tissues, as well as 13 500 hexahedral elements describing the flat support. Simulations were set up using the PreView module of FEBio (Ver. 1.18), analysed using the Pardiso linear solver of FEBio (http://mrl.sci.utah.edu/software/febio) (version 2.3.1), and post-processed using PostView of FEBio (version 1.9.0).[40]

We compared distributions of the SED and maximal shear strains in muscle, fat, and skin tissues during sitting, in a volume of interest (VOI), including all of the soft tissues elements below the imaginary horizontal line connecting the lowest points of the heads of the femurs.

3 | RESULTS

The stress-strain curves of the stacks of porcine skin that were tested under uniaxial compression are shown in Figure 3. The effective compressive response of the porcine skin is typical for highly viscoelastic non-linear materials subjected to quasi-static compression. Calculated elastic moduli ranged between 42.8 (kPa) and 92.58 (kPa), which implies that the porcine skin specimens that were used are indeed representative of aged human skin in terms of stiffness.[41]

The COFs of porcine skin when rubbing against the different medical textiles that were tested, in the aforementioned dry and wet conditions, are shown in Figure 4. The COFs of dry skin, when rubbing against the tested fabrics,

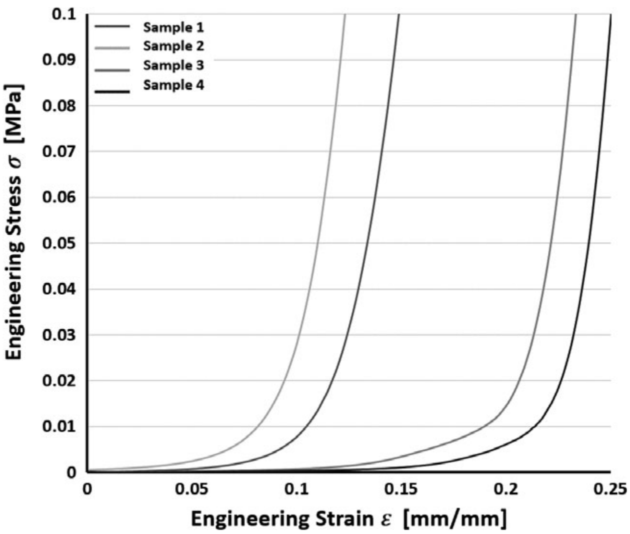

FIGURE 3 Stress–strain curves of the 4 porcine skin samples that were experimentally tested in uniaxial unconfined compression

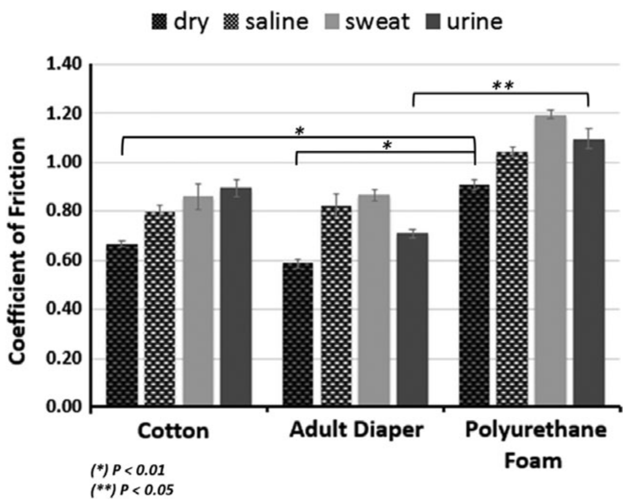

FIGURE 4 Comparisons of the coefficients of friction of porcine skin, dry and wet, with saline, sweat, or urine, rubbing against a standard cotton hospital bed sheet, a standard adult diaper, or a standard polyurethane foam dressing. Results are expressed as means ±1 SD

varied between 0.59 (standard adult diaper) and 0.91 (standard polyurethane foam). In addition, the COF consistently increased with the addition of saline, sweat, or urine compared with the reference (dry) skin conditions, regardless of the textile that was tested. We found a significant difference between the COFs resulting from dry skin and skin that became wet by sweat or saline ($P < .05$); however, the use of urine did not yield significant difference from the dry skin cases. We further found statistically significant differences between COFs measured between dry skin and cotton sheet or the cotton-like synthetic layer of the adult diaper and the polyurethane dressing, as well as between skin wet with urine, rubbing against the adult diaper and the polyurethane dressing (Figure 4).

SED distributions in the soft tissues of the buttocks, under the influence of high and low COFs, are shown in Figure 5A. Peak SED values were found at the bone-fat interface for both cases of high and low COFs; however, when high COF was used, the volumetric exposures of the soft tissues to elevated SED values increased considerably. Specifically, the volume of soft tissues exposed to SED values exceeding 0.5 and 1.5 kPa increased by 12% and 52%, respectively. An example time course of the developing distributions of maximal shear strains during the sitting-down process, when high COF is used, is shown in Figure 5B. Peak maximal shear strains were reached when full weight bearing is achieved; strains were maximised in the deep muscle tissues, adjacent to the bone-muscle interface. This last finding is highly important and should be highlighted as it demonstrates that a rise in skin COF, fir example, because of moisture and wetness, will cause elevated deep tissue deformations (and not only high skin deformations) when shear forces apply.

4 | DISCUSSION

In this work, we investigated the effects of moisture caused by human sweat, urine, or saline on the resulting COF of porcine skin against different textiles used in the medical setting in the context of PI prevention, which is now the focus of attention of health care organisations, clinicians, academics, and industry. We performed physical measurements of static COFs using a tilting-table tribometer followed by FE computational modelling in order to illustrate the effect of increased COF at the skin on strains and stresses that develop deep within the soft tissues of the buttocks.

Our COF measurements, both under dry and wet skin conditions, were performed using the well-established tilt method.[9,42–48] Statistically significant differences were found between the dry skin cases (against all of the textiles) and the cases of moisture because of sweat and saline. These results were probably obtained because of the relatively high mineral content in saline and sweat. Specifically, the sweat that we used was collected from athletes during the first 30 minutes of practice and was hence characterised by a higher mineral content compared with subsequent sweat samples.[49] When water evaporates from the skin surface, the remaining salts generate a thin textured layer, which contributes to a higher COF compared with liquids with lower mineral content.[50] The insignificance that was found when urine was tested (compared with the dry cases) can be attributed to the limited number of measurements as the trend of increase in COF when any type of moisture is added is clear. The increase of the COF because of wet conditions can be attributed to the increase in the adhesive bonds, which develop between the skin and the fabrics.[9–11,16,17] The presence of liquids at the skin interface creates a plasticizing effect—smoothening of skin roughness and, consequently, an increase in the effective skin-fabric contact area, which elevates the adhesion component of friction. Hence, moisture

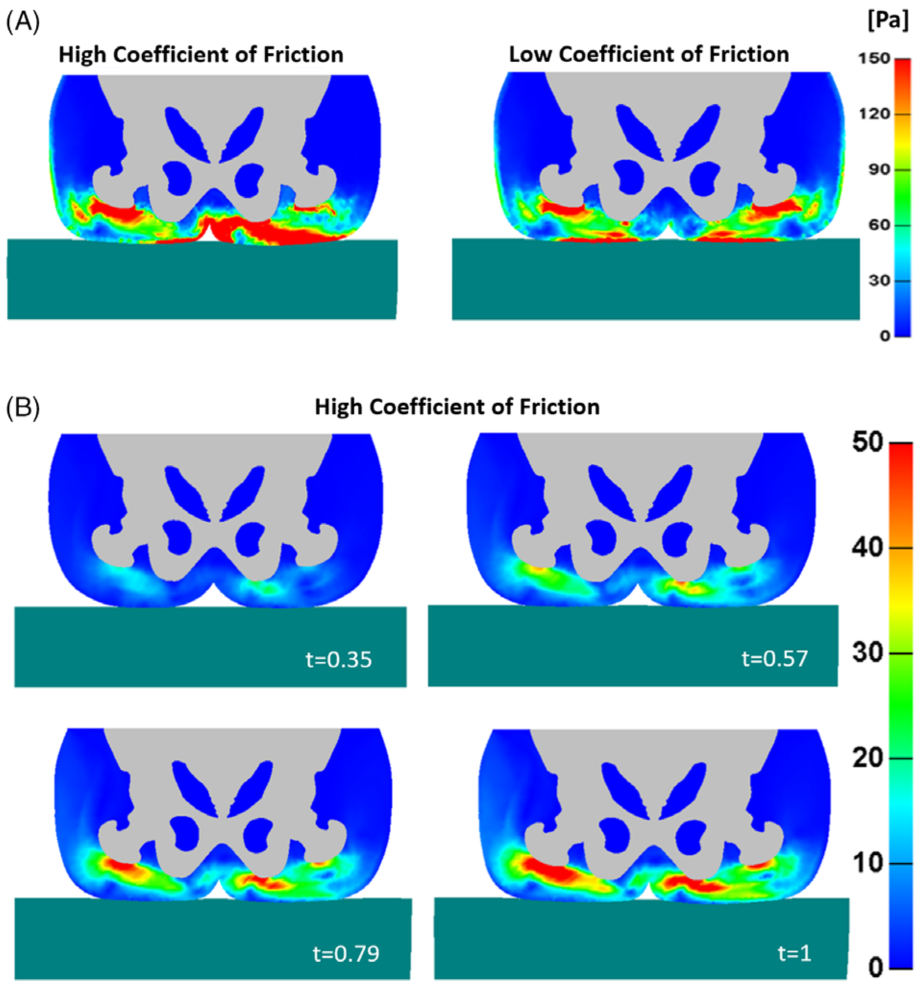

FIGURE 5 A, Distributions of strain energy density (SED) in the soft tissues of the buttocks, with the use of a high coefficient of friction (1.2, left) and a low coefficient of friction (0.6, right). B, An example time course of the distributions of shear strains, which develop under the influence of high coefficient of friction during sitting, from the time of initial skin-cushion contact ($t = 0$) to full weight bearing ($t = 1$)

and wetness can, in fact, accelerate and promote the formation of PIs.

In addition, statistically significant differences were identified between COFs measured between dry skin and a cotton sheet or an adult diaper compared with the case of polyurethane foam dressing. As opposed to the fibrous structure of the cotton and cotton-like synthetic layer of the adult diaper, polyurethane foams are characterised by a porous structure and rough surface. These characteristics result in an increased contact area between the foam and the skin and, possibly, also cause increased accumulation of liquids at the interface and, as a result, lead to higher adhesion forces and greater COFs.[7]

Comparing our results with a recent study by Vilhena and Ramalho,[51] the COFs measured in our study (for dry fabrics), which are almost twice as high, can be explained by the different measurement methods. Vilhena and Ramalho used a method where the normal force is controlled, and the friction force is measured using horizontal sliding. In the tilt method, however, the static COF is estimated by tan(θ). The tangent function is non-linear and increases at a greater rate for large values of θ than for small values. For a static COF greater than 1.0, the inclination angle (θ) is greater than 45°, which makes the tilting-table method more sensitive to measurement errors when measuring greater inclination angles, corresponding to greater COFs.

In our FE simulations, we utilised a geometrical illustration of the buttocks during the sitting-down process, with high versus low COFs (Figure 5), in order to illustrate how changes in the COF of skin affect internal shear loads that develop deep within the weight-bearing tissues of the buttocks. We found that the reduction of friction at the skin-textile interface is a key measure in the prevention of skin, as well as DTIs, as low COF at the interface between the skin and the textile cause considerably lower exposures of tissues to elevated SED values and shear loads. In the literature, SED distributions have been experimentally correlated with the severity and extent of tissue damage.[39] In addition, it is a well-accepted scalar measure for quantifying internal tissue loads as related to PIs and for practical purposes in evaluating the effects of medical materials and devices that come into contact with the body of a patient.[37] We thus opted to use the SED measure in order to compare the response of the soft tissues of the buttocks to high and low COFs during sitting. We further examined the distributions of maximal shear strains in the soft tissues of the buttocks during sitting as a second measure of the risk of PIs. Our results demonstrated that, during the sitting-down process, under the influence of a high COF (eg, representative of wet conditions), elevated shear strains develop deep within the soft tissues, at the bone-muscle interface, thus jeopardising the viability of deep soft tissues, not only of skin.

In conclusion, we demonstrate that moisture is, in fact, a decisive physical factor in the development of PIs and may accelerate their formation by increasing the COF between the skin and medical fabrics, regardless of the type of the liquid that is present. Hence, reduction of the wetness/moisture between the skin and fabrics in patients at a high risk of developing PIs is a key measure in PI prevention. This can be achieved by ensuring the skin-fabric contact area remains clean and dry; frequent replacement of bed sheets, diapers, and dressings; and by performing frequent and careful repositioning in order to minimise shear loads.[52] Furthermore, the use of smooth and less fibrous textiles should reduce the friction at the skin-fabric interface, reduce the shear loads delivered to the deep tissues, and overall lower the risk of PI development.[53]

ORCID

Amit Gefen https://orcid.org/0000-0002-0223-7218

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