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Moisture-Responsive Friction Adaptability: Rethinking the Conventional Skin Silicone Interfaces in Pressure Injury Prevention Dressing Designs

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ABSTRACT

This study evaluated the frictional properties of the skin-contact interfaces of two multilayer prophylactic dressings under simulated perspiration conditions. The tested dressings were identical except for the skin-contact interface, which was either silicone-made or Hydrofiber-made, that is, AQUACEL Hydrofiber Technology. Using a standardised tribological ‘sled test’ setup and a skin-mimicking polymer substrate, we measured the static and kinetic coefficients of friction for each dressing interface type at varying moisture levels. The dressing with the Hydrofiber interface consistently demonstrated significantly lower static and kinetic coefficients of friction compared to the silicone-based dressing, across all moisture conditions. The Hydrofiber interface exhibited a sharp coefficient of friction reduction with minimal (5%) moisture accumulation, mimicking overnight perspiration under thermoneutral conditions. This dressing maintained the low coefficient of friction levels at a steady level of approximately 0.2 until full saturation. In contrast, the silicone interface retained high (> 1) coefficients of friction regardless of moisture. These findings highlight an important biomechanical advantage of Hydrofiber skin-contact materials in reducing frictional forces at the skin-dressing interface, especially in moisture-prone body areas, in a pressure injury prevention context. Friction-responsive skin-contacting dressing materials with low coefficients of friction, which remain low while they become moist due to perspiration accumulation, should be preferred for preventative dressings.

1 | Introduction

Pressure injuries (PIs) are a persistent clinical challenge, affecting vulnerable patient populations worldwide and contributing to increased morbidity, extended hospital stays and substantial healthcare costs [1]. A key biomechanical factor contributing to the development of PIs is soft tissue deformations caused by shear forces, which occur at the interface between the skin and any contacting support surfaces [2, 3]. Multilayer dressings are commonly employed in pressure injury prevention (PIP)

protocols. However, their effectiveness is influenced not only by their capacity to alleviate the soft tissue stresses under the dressing and manage moisture and the microclimate, but also by their static and dynamic frictional interactions with the skin [4, 5]. Biomechanical evidence suggests that the coefficient of friction (COF) between the skin and a dressing (or any other clinically relevant surface or object) is critical in determining the magnitude of internal tissue shear during movements such as sliding-in-bed, repositioning, transfers, or micro-movements on a support surface [6] (Figure 1). Importantly, many commercially

Abbreviations: AHT, AQUACEL Hydrofiber Technology; COF, coefficient of friction; HOB, head-of-bed; PIP, pressure injury prevention; PIs, pressure injuries.

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Key Messages

- A Hydrofiber skin-dressing interface reduces friction more than a silicone one.
- Moisture sharply lowers friction for Hydrofiber but not for silicone dressings.
- Low coefficient of friction reduces soft tissue shear and pressure injury risk.
- Dressings with Hydrofiber interface adapt better to perspiration-prone areas.
- Dressing material choices impact the friction and injury prevention efficacy.

available prophylactic dressings feature an adhesive silicone skin-contact layer that was originally designed for wound treatment applications, where adherence (stay-in-place) and atraumatic removals are prioritised. Indeed, silicone was consistently reported to have high COFs with skin [8, 9]. Accordingly, when these silicone interfaces are used in a PIP context, they may paradoxically transmit the external frictional loads at the skin-dressing interface inward, directly into susceptible tissues under bony prominences which are substantially distorted, thereby increasing, rather than decreasing, the internal tissue shear stresses.

Only a few studies have measured the frictional properties of dressings. Call et al. studied the frictional properties of dressings applied to a gel simulating skin, against foam mounted on a sled [10], whereas Ohura et al. mounted dressings to porcine skin and reported that the dressings reduced both the normal and shear forces on the skin during tangential displacements [11]. Sprigle et al. quantified the static and kinetic COFs of prophylactic dressings using a tribometer while the dressings were in contact with bed linen materials [12]. Interestingly, all these studies focused on the frictional properties at the external surface of the dressing (which faces the bedlinen or the support surface) and overlooked the other critically important interface, namely, at the skin-dressing side. The only study focusing on the skin-dressing interface was published by Grigatti and Gefen [13], who used a tilting-table tribometer to study the COF of a hydrogel dressing against fresh, shaved porcine skin to characterise the tacky behaviour of this hydrogel-based dressing. The latter work also considered perspiration from the skin by uniformly spraying saline solution onto the porcine skin prior to the COF measurements, highlighting the importance of considering the sweat response in the development of clinically relevant test protocols in the context of PIP by means of dressings [13].

The skin microclimate, and in particular, the presence of perspiration or moisture at the skin-dressing interface further impacts the COFs, and hence, the intensity of the frictional forces acting on the skin, which are proportional to the COF levels [6, 7, 14–16].

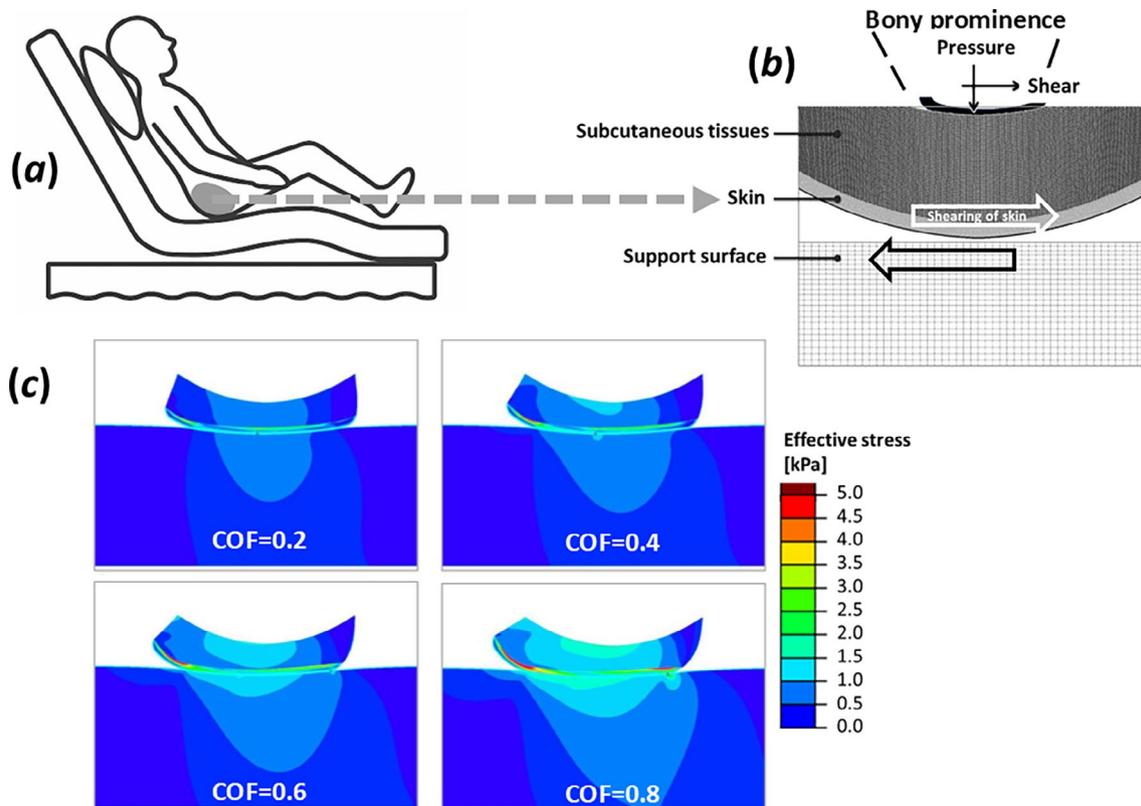


FIGURE 1 | Biomechanical computational modelling of a skin region under a bony prominence in a person lying supine in a hospital bed, to demonstrate the effect of the coefficient of friction (COF) at the skin-support interface on the internal soft tissue stress state. As gravity pulls the body of the patient towards the foot of the bed (a), skin and underlying soft tissues are sheared between the support surface and the bony prominence (b), and the intensity of the forming stress concentrations rises with the COF value at the moving (skin-support) interface (c). *Source:* Reproduced from Shaked and Gefen [7] under the Creative Commons Attribution Licence (CC BY).

This implies that these COFs are also influenced by the ambient conditions, that is, the room temperature and relative humidity, and also by the core body temperature and skin temperature of the patient [14]. In theory, moisture accumulation can either reduce or increase (or insignificantly affect) the COFs, and therefore, the frictional forces and soft tissue stresses, depending on the dressing material which is in contact with skin. Despite this, limited data exist in the literature comparing COFs of different skin-interfacing dressing materials under varying moisture conditions, using clinically-relevant laboratory models.

This in vitro study aimed to address this knowledge gap by evaluating the static and kinetic COFs of two multilayer dressing types with distinct skin-contact materials: A silicone-based interface and a Hydrofiber-based interface, that is, AQUACEL Hydrofiber Technology (AHT). The COFs of these skin-contact materials were measured in a bioengineering laboratory setup under controlled, simulated perspiration conditions. By systematically assessing how incremental moisture accumulation affects the COFs at a model of a moving (frictionally-sliding) skin-dressing interface, this study provides critical insights into the biomechanical suitability of skin-contacting dressing materials intended specifically for PIP.

2 | Methods

This laboratory study was designed to evaluate the COF of multilayered wound dressings at a simulated skin interface under conditions representing varying levels of perspiration. Two commercially available dressing types were assessed: One featuring a silicone contact layer with skin (Dressing A) and another with an AHT¹ contact layer (Dressing B). Dressings A and B were identical in all aspects, that is, in their shape and size, construct design and material components, excluding the skin-interfacing layer as noted above. The COF measurements were performed based on the ASTM D1894-14 test standard [17], using a Zwick material testing machine equipped with a 100N load cell (pre-calibrated according to manufacturer guidelines) and a dedicated COF sled assembly (Figure 2). The setup included a supporting base, a precision glass plane (200 mm × 200 mm × 4 mm thick) and a pulley system connected to the load cell via a beaded chain (spring rate ≥ 10.7 kg/mm). A silicone elastomer skin simulant, Dragon Skin (Smooth-On Inc., Macungie, Pennsylvania, USA)

which is well-accepted in wound care research [18, 19], was cast to a thickness of 3.5 mm and affixed to the test plane (Figure 2). This surface served as a consistent and repeatable surrogate for human skin for all dressing tests.

Square samples (63.5 mm × 63.5 mm) of each dressing type were prepared. For Dressing A (silicone interface), which displayed a higher frictional interaction as detailed further in the Results section, an additional 500 g of weight was applied atop the sled, based on the outcomes of preliminary testing, to minimise slipping and ensure a stable contact.² For Dressing B (AHT interface), a total sled weight of 200 g was used. Dressings were applied with their respective contact layers (silicone or AHT) in full contact with the skin simulant. Special care was taken to prevent alteration of the contact surface during sample mounting. To model clinical sweat conditions, 0.9% isotonic saline (9 g of NaCl dissolved in 1 L of distilled water) was used as a perspiration simulant and applied to the dressing specimens in 1.3-g increments through pipetting directly onto the dressing, following a previously reported sweat simulation protocol [20]. The moisture levels tested included 0% (dry), 5%, 10% and 100% (full) saturation. Each dressing sample was weighed pre- and post-application to ensure accurate moisture quantification, and testing of moist dressings began immediately thereafter. The ambient temperature of the laboratory was approximately 20°C, and the relative humidity was approximately 50%.

The sled was connected to the load cell via the beaded chain and dragged across the skin simulant surface at a constant rate of 150 mm/min (Figure 2). The static COF was determined by the maximum initial force required to initiate the sled movement, while the kinetic COF was calculated as the average force during sustained movement. Each dressing specimen was tested in both the machine and transverse directions (i.e., sled placement and testing were repeated after rotating the sled by 90°) to account for and minimise potential directional variability. The output from the testing machine was digitally captured and processed using the 'testXpert III' software provided by the manufacturer of the material testing machine. Each condition was replicated at least six times per dressing type. Results were expressed as mean ± standard deviation. Statistical comparisons between dressing types and moisture levels were conducted using analysis of variance (ANOVA) followed by Tukey–Kramer multiple pairwise comparisons, with the level of statistical significance set at $p < 0.01$.

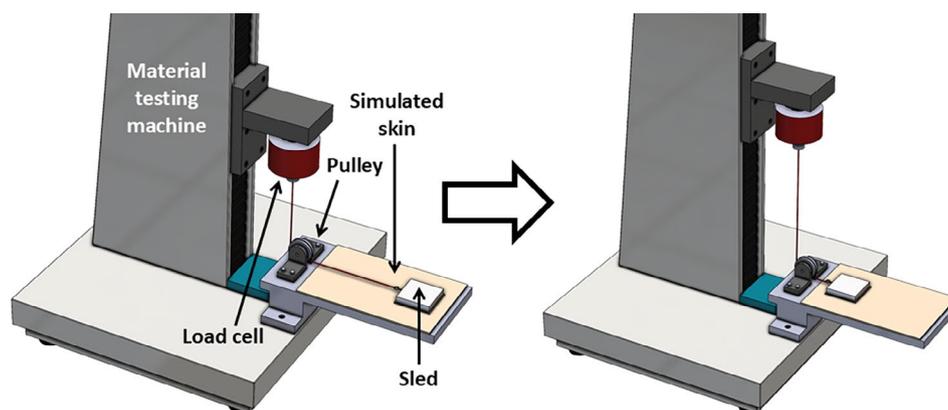


FIGURE 2 | Setup of the tribological sled test apparatus used for the coefficient of friction measurements. Dressings were adhered to the bottom of the sled (left frame) and dragged over the skin simulant (right frame) at a constant speed (150 mm/min) by means of the material testing machine.

3 | Results

Regardless of the moisture level, Dressing B (AHT) consistently demonstrated statistically significantly lower static and kinetic COF values compared to Dressing A (silicone), indicating superior friction-reducing properties of Dressing B ($p < 0.01$, Figure 3, Table 1). Specifically, under dry conditions (0% moisture), Dressing B exhibited approximately a 2.8-fold lower static COF (1.55 ± 0.45) compared to Dressing A (4.38 ± 0.58), and a 1.8-fold lower kinetic COF (1.41 ± 0.09 for Dressing B versus 2.59 ± 0.87 for Dressing A). Importantly, Dressing B demonstrated a sharp, nearly immediate reduction in its COF upon minimal (5%) moisture accumulation (Figure 4). Specifically, at just 5% saturation, both the static and kinetic COFs of Dressing B dropped significantly ($p < 0.01$) by approximately three-fold, to 0.52 ± 0.15 and 0.47 ± 0.05 , respectively (Figure 4 and Table 1). These reduced COF levels persisted consistently even at full saturation (100% moisture), with minimal additional reduction observed beyond the 10% moisture level (Figure 4 and Table 1), indicating a threshold effect of moisture on the frictional properties of the AHT interface. In contrast, Dressing A retained relatively high COF values when fully saturated (COF at 100% moisture: static 1.09 ± 0.23 ; kinetic 1.47 ± 0.35).

4 | Discussion

This laboratory study builds on our prior published works, highlighting the importance of minimising frictional forces at the skin surface [21–23], and particularly, when using a prophylactic dressing, at the skin-dressing interface [13], to reduce the risk of PIs in at-risk patients. The COF at the skin-dressing interface plays a

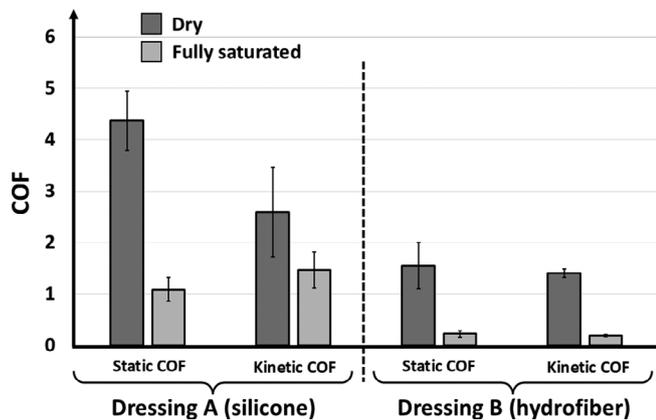


FIGURE 3 | The coefficient of friction (COF) data for the two types of dressings investigated in this study, in their dry and fully saturated states.

TABLE 1 | Coefficient of friction (COF) comparisons for the two studied dressing types.

Moisture level (%)	Dressing A		Dressing B	
	Static COF	Kinetic COF	Static COF	Kinetic COF
0	4.38 ± 0.58	2.59 ± 0.87	1.55 ± 0.45	1.41 ± 0.09
5	N/A	N/A	0.52 ± 0.15	0.47 ± 0.05
100	1.09 ± 0.23	1.47 ± 0.35	0.22 ± 0.06	0.19 ± 0.02

Note: Each test condition was replicated at least six times per dressing type, and results were expressed as mean \pm standard deviation. Of note, the calculation of COFs accounts for the differences in sled weights.

pivotal role in generating internal soft tissue shear stresses, which are widely recognised as a critical biomechanical factor in PI aetiology [23–27]. Our current findings demonstrate that a multilayered dressing with an AHT skin-contact layer, namely, Dressing B, consistently exhibited statistically significantly lower static and kinetic COFs compared to a silicone-based dressing (Dressing A) across all moisture levels tested (Figure 3 and Table 1).

Previous research has shown that high interfacial frictional forces at the skin surface leads to elevated shear stresses in skin and subcutaneous tissues, which can potentiate the cascade of cell and tissue damage under sustained loading, especially when combined with moisture at the interface and/or impaired tissue perfusion [6, 7, 21–27]. Silicone-based skin contact layers are commonly used in PIP dressings due to the atraumatic removal properties of silicone [28], given that the vast majority of commercial PIP dressings were originally developed for treatment of existing wounds where atraumatic removals are critical. A silicone-based interface retains relatively high friction (i.e., COFs > 1) with skin at either dry or moist conditions (Table 1 and Figure 3), as it was originally designed to be ‘sticky’ (tacky) in order to maintain the dressing in place. However, while transitioning the silicone-foam dressing technology from treatment to preventative applications, the question of whether the silicone interface potentially transmits frictional loads deeper into vulnerable soft tissues when the body and soft tissues of a patient

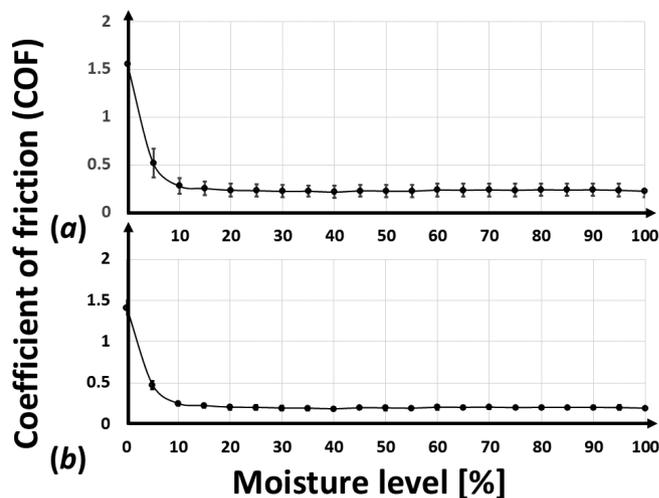


FIGURE 4 | The drop in the static coefficient of friction (COF) values of Dressing B with the increase in the moisture level within this dressing. A sharp and statistically significant drop of the COF occurred already between the dry and the minimal 5% moisture level for both the static (a) and kinetic (b) COFs of Dressing B, and these COF data plateaued thereafter for all the greater moisture levels.

are experiencing shear forces and deformations (e.g., while sliding in bed when the head of the bed is elevated; Figure 1) was surprisingly never introduced in the literature. Indeed, the present results confirm, highlight and extend these concerns. Dressing A with its silicone skin interface maintained high COF values, above unity, under all test (dry and moist) conditions, as expected and precisely as it was designed to function (on both dry and moist skin). In contrast, Dressing B with the AHT skin interface demonstrated a marked reduction in both static and kinetic COFs with minimal moisture accumulation (COFs of approximately 0.5 with as little as 5% saturation; Table 1 and Figure 4), and these low values further halved and persisted for 10% moisture, indicating a threshold effect between dry and 10%, that is, at approximately 5% moisture level (Figure 4). This ‘threshold effect’ is especially clinically relevant in low-sweat anatomical regions, where even modest trans-epidermal water loss could be sufficient to trigger these COF reductions while an AHT-based interface is used, thereby reducing the internal soft tissue shear exposures within a relatively short time post-application of these dressings.

These laboratory findings align with prior investigations that emphasised the role of microclimate modulation and energy dissipation within dressings as integral to successful PIP performance [20]. Specifically, it was reported that under resting, thermoneutral, overnight conditions, silicone-foam dressings accumulate approximately 0.3 g of sweat [20], which indicates that for an AHT-interface dressing with an assumed dry weight of 8 g, the ‘threshold effect’ of reduction of the COFs (Figure 4) will occur already after a first night of use. For dressings where the intended PIP usage time is 5–7 days (which is a common instruction for use in a PIP context), the implication is that the AHT dressing functions with a COF that is at least approximately half of silicone-interface dressings on the first day. Moreover, the COF of the AHT dressing will likely be more than five times lower than that of the silicone-foam dressings during the four to six additional days. Accordingly, while COFs alone should not be the sole determinant in PIP dressing selection, their influence on shear mitigation should not be overlooked. The current data advocate for questioning whether silicone should be the contact layers in a PIP dressing (as opposed to treatment dressings), because in a PIP dressing, more friction-responsive and adaptive solutions such as AHT technologically exist, and are commercially applicable, and can be highly beneficial especially in moisture-prone clinical environments such as the sacrum or heels.

Some limitations of this *in vitro* study should be acknowledged. First, the use of a silicone skin simulant (Dragon Skin) provided a standardised and reproducible surface for the tribological studies and for measuring the frictional properties in interaction with the dressings, but it does not fully replicate the complex anatomical, physiological and biomechanical properties of living skin. This includes the influences of aging, skin roughness and wrinkling and skin diseases or chronic diseases affecting skin behaviour and properties and potentially the COF of skin in interaction with other materials, such as diabetes [29]. Additionally, the experimental design involving unidirectional sliding under constant speed and loading simplifies the multifactorial nature of *in vivo* frictional forces acting on skin, and the potential interactions with a changing microclimate. Future

studies may benefit from incorporating anatomically relevant robotic platforms that better mimic real-world clinical scenarios [30]. Furthermore, the simulated skin surface did not represent hair, though the presence of hair is known to affect the COF of skin with contacting materials [15, 16]. Likewise, variations in skin roughness, wrinkling, age-related epidermal thinning or disease-related skin alterations such as those associated with diabetes were not considered in the present work. These factors may further modulate interfacial friction, either directly or indirectly by increasing skin fragility and susceptibility to surface defects or tears. Such micro- or macro-scale skin irregularities could potentially shift the moisture level at which the observed threshold reduction in friction occurs for the AHT interface. However, because compromised skin integrity is generally associated with elevated baseline skin friction [6, 15, 22], the moisture-responsive friction reduction observed for the AHT interface may be particularly relevant in these populations, meriting future condition-specific investigations. With regard to long-term performance, friction was measured immediately following moisture application. However, AHT interfaces are designed to remain in a gelled, hydrated state once fluid is absorbed. Therefore, the low-friction regime observed here is expected to persist during continued clinical use, although direct long-term tribological validation is warranted. Lastly, although different sled weights were used to ensure stable contact for the silicone and AHT interfaces, the load cell sensitivity and resolution were sufficient to ensure accurate force measurements under both loading conditions, and the calculated COFs are independent of the applied normal force in accordance with ASTM D1894.

Patient migration in bed commonly occurs following head-of-bed (HOB) elevation. Motion-capture studies have shown that patients can migrate on the order of approximately 4–10 cm during HOB elevation to 45°–65°, indicating substantial relative motion occurring over a short time window associated with the elevation manoeuvre [25]. This initial migration may be followed by continued, gradual, gravity-driven movement occurring over longer periods, such as during a standard 2-h repositioning interval. For context, a midrange migration distance of approximately 7 cm occurring during HOB elevation over approximately 10–20 s corresponds to an average relative sliding speed of ~210–420 mm/min, whereas the same displacement occurring gradually over a standard 2-h repositioning interval corresponds to ~1 mm/min. Thus, clinically relevant skin-surface motion spans orders of magnitude in velocity, and the sliding rate used in the present study (150 mm/min) lies within this clinically relevant range while enabling standardised comparison of kinetic COF behaviours between the studied dressing interfaces.

In summary, the superior frictional performance of the AHT-interface dressing (dressing B), and in particular, its responsiveness to physiologic moisture levels generated by spontaneous sweat under thermoneutral, resting conditions, suggests that AHT interfaces with skin may offer better protection against shear-induced soft tissue distortions which are associated with PIs in PIP dressings. This bioengineering laboratory research and the insights obtained further emphasise the need for holistic evaluation of dressings for PIP applications in terms of their ability to attenuate frictional forces and alleviate soft

tissue stresses. In particular, factoring in the skin-dressing COF properties alongside how these might be affected by moisture accumulation and the microclimate conditions in the relevant clinical setting is a key consideration.

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Ethics Statement

The authors have nothing to report.

Consent

The authors have nothing to report.

Conflicts of Interest

The authors declare no conflicts of interest.

Data Availability Statement

The data that support the findings of this study are available from the corresponding author upon reasonable request.

Endnotes

¹AQUACEL Hydrofiber Technology is a proprietary Convatec Ltd. (Deeside, UK) technology of soft material made for direct contact with skin or wound tissues, composed of sodium carboxymethylcellulose fibres that gel upon exposure to moisture.

²While sled weights differed to ensure stable contact and prevent premature slipping, these adjustments do not affect the comparability of measured COF values, as the COF is normalised to the applied normal force and thus, by definition, it remains independent of the normal forces under steady-state contact conditions.

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