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Modelling and Simulation of Soft Tissue Loading Reduction at the Posterior Heel by a Moisture-Responsive Prophylactic Dressing

Daria Orlova[1] | Amit Gefen[1,2,3,4]

[1]School of Biomedical Engineering, Faculty of Engineering, Tel Aviv University, Tel Aviv, Israel | [2]Skin Integrity Research Group (SKINT), University Centre for Nursing and Midwifery, Department of Public Health and Primary Care, Ghent University, Ghent, Belgium | [3]Department of Mathematics and Statistics and the Data Science Institute, Faculty of Sciences, Hasselt University, Hasselt, Belgium | [4]Susan Wakil School of Nursing and Midwifery, Faculty of Medicine and Health, University of Sydney, Sydney, Australia

Correspondence: Amit Gefen (gefen@tauex.tau.ac.il)

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ABSTRACT

Pressure ulcers/injuries at the posterior heel commonly develop in patients with limited mobility or sensation, where movements such as sliding in bed generate shear forces that elevate the deep soft tissue stresses. Prophylactic multilayer dressings are recommended to reduce this injury risk, yet many current products incorporate high-friction silicone adhesives for attaching the dressing to the skin, which may increase shear transfer to the skin and subdermally while a patient is moving or being moved. This study evaluated a multilayer dressing featuring an Aquacel Hydrofiber Technology skin-contact layer that exhibits moisture-responsive friction reduction. Finite element models of a posterior heel resting on a support surface were developed to simulate compressive and shear loading during patient sliding-in-bed, comparing no dressing, dry (new) and moist (used) dressing conditions. Moisture exposure reduced the skin-dressing coefficient of friction from 1.55 to 0.22, leading to substantial reduction in soft tissue strain and stress exposures, such that high-magnitude strain and stress concentrations (> 75th-percentile) were effectively absent within the modelled loading domain under the simulated conditions. Moisture-induced friction reduction facilitated controlled micro-sliding at the skin interface, attenuating shear transfer into skin and subcutaneous tissues. These findings demonstrate that moisture-responsive interfacial friction adaptation is an important mechanism of action contributing to the biomechanical protective efficacy of prophylactic dressings.

1 | Introduction

Pressure ulcers/injuries (PUs/PIs) remain a major healthcare concern globally, particularly among patients with impaired mobility or sensation. The posterior heel is especially vulnerable due to thin soft tissues and concentrated loading in the supine position [1, 2]. Heel PUs are particularly common in mechanically ventilated patients,[1] where head-of-bed elevation promotes sliding in bed (also known as migration in bed) and causes greater perpendicular forces on the calcanei due to offloading of the shoulders. The sliding in bed, combined with the increased bodyweight force fraction on the feet, elevates the shear loading intensity on the posterior heels [3–9]. This high shear loading interacts with the anatomically thin heel tissues, rapidly elevating the local, internal soft tissue stresses that predispose to a deep tissue injury,

Abbreviations: AHT, Aquacel Hydrofiber Technology; ANOVA, analysis of variance; COF, coefficient of friction; FE, finite element; PEI, protective efficacy index; PI, pressure injury; PU, pressure ulcer; ROI, region of interest.

Summary

- Finite element modelling simulated heel soft tissue loading during bed sliding.
- Mild moisture reduced the skin-dressing friction coefficient nearly eightfold.
- The moisture exposure consequently lowered stress concentrations in soft tissues.
- The high-magnitude strain and stress concentrations were effectively absent.
- A moisture-responsive friction-reducing skin interface layer contributes to protection.

resulting in extended hospitalisation and increased mortality. The heel is also a common anatomical site for prophylactic dressing application in clinical practice and a frequent target for preventive interventions because of its high susceptibility to PU development. Accordingly, recent international clinical guidelines recommend applying prophylactic dressings over high-risk body regions including the heels to alleviate the soft tissue loading and mitigate the PU risk [10, 11]. However, the suitability of dressing technologies originally developed for therapeutic wound management for prophylactic use on intact skin, including whether added design complexity translates into meaningful preventive benefit, remains incompletely understood.

The current (2025) international guideline for PU/PI prevention and treatment conditionally recommends applying 'soft silicone adhesive multilayered foam' dressings to the heels [11]. However, guideline recommendations are informed by multiple considerations (e.g., certainty of evidence, feasibility, costs, equity and implementation factors) and therefore do not imply equivalence among all dressings within a recommended category. The guideline does not specify design criteria or mechanisms of action for these dressings, and important design differences between products that may affect performance have been reported [12–15]. In this context, the basis for specifying silicone adhesive interfaces in the most recent edition of the guideline [11] is not explicitly linked to the mechanisms through which they may contribute to protective effects when used prophylactically. Although silicone adhesive interfaces are widely used to secure therapeutic dressings, including when a patient is moving or during showers, their high-friction contact is not necessarily advantageous for prophylactic use, where minimising shear transfer to the skin and underlying tissues is critical. Importantly, the question of whether high-friction silicone interfaces provide superior or inferior protection by an applied dressing relative to low-friction alternatives has not been directly addressed in the literature in the context of PU prevention.

Multilayer dressings with alternating layer stiffnesses can dissipate shear forces and reduce soft tissue strains and stresses through deformations of the softer layers in-between stiffer layers; a more recently identified mechanism of action involves frictional sliding between adjacent dressing layers, which further increases the effectiveness of internal shear absorption in the dressing [1, 13, 16–20]. However, these types of dressings intended for prophylactic use incorporate a silicone contact layer with skin, which induces a high coefficient of friction (COF) at the skin-dressing interface [21–23]. In contrast, a low COF reduces the transfer of shear forces into the skin and underlying tissues. Indeed, in general PU prevention practices, minimising frictional forces at the skin-bedsheet or skin-clothing interfaces is well established and desirable to limit shear transmission into skin and subdermal tissues, and specialised low-friction fabrics and textiles for medical use have been developed for these purposes [24–26]. Similarly, external surfaces of dressings typically have low COFs to support shear reduction, by limiting transmission of frictional forces into the dressing and from there, to underlying soft tissues [27]. Thus, the prevailing assumption that silicone-adhesive-induced high friction at the skin-dressing interface is desirable may be biomechanically counterproductive in PU prevention contexts. This is because a fully adhesive interface constrains micro-sliding and may increase shear stress transmission to the skin and subdermal tissues.

The frictional behaviour of the skin-dressing interface can, theoretically, be modulated by moisture, as certain advanced materials suitable for wound care, e.g., hydrogels originally developed for cartilage replacement, may reduce their COF when hydrated [28, 29], potentially enabling controlled micro-sliding at the interface that reduces frictional force transmission [30]. Comparable moisture-responsive friction reduction has also been demonstrated in consumer textiles to reduce chafing, again demonstrating the general biomechanical principle that hydration can be used to modulate the interfacial friction [31, 32]. However, in the context of prophylactic dressings, the quantitative biomechanical implications of friction adaptability on tissue protection have not yet been examined. The present study addresses this gap by integrating COF test data obtained using a validated method with advanced computational modelling to evaluate the biomechanical efficacy of a commercial multilayer prophylactic dressing with non-silicone skin-interfacing layer that exhibits a moisture-responsive friction adaptation [30]. This allowed evaluation of how moisture exposure alters its protective function under shear loading.

Using an anatomical model of the heel resting on a support surface, we simulated soft tissue loading during patient migration in bed, providing the first quantitative assessment of how moisture-induced reductions in the dressing-skin COF influence the soft tissue loading state. By elucidating the moisture-responsive friction-adaptation mechanism, the study advances understanding of how dressing material behaviour under physiological hydration can contribute to preventive protection against PUs. We hypothesised that moisture-induced reduction in the skin-dressing COF would decrease the internal soft tissue strains and stresses in the posterior heel during shear-loading conditions. The objective of this study was to quantify the effect of moisture-induced COF reduction on heel soft tissue loading using computer modelling informed by experimental measurements. This is the first study to quantitatively evaluate how moisture-induced reduction in skin-dressing COF influences the internal soft tissue stress concentrations associated with a heel PU risk level.

2 | Methods

2.1 | Geometry and Mechanical Properties

Two FE model variants of the posterior left heel resting on a flat medical foam support surface were developed using the anatomically accurate, MRI-based modelling framework described in detail in our previous work [1]. The two variants differed only in the application of the studied multilayer dressing (ConvaFoam Border, Convatec Ltd., Deeside, United Kingdom): one variant without the dressing and one with it applied prophylactically. The underlying heel anatomy was identical in both variants and included the posterior calcaneus, Achilles tendon, adipose tissue, and skin (Figure 1). The modelled applied dressing consists of four distinct layers, namely an Aquacel Hydrofiber Technology layer (AHT, layer I), a superabsorbent fibre layer (layer II), a soft cushioning foam layer (layer III) and an outer protective film (layer IV). The inner three layers I, II, and III, which are subject to frictional interactions, were explicitly modelled with physical thicknesses of 2, 2.5, and 2.5 mm, respectively. The outer protective film layer (layer IV), which is positioned against the support surface, is substantially thinner and was not modelled as a separate deformable body due to its negligible mechanical thickness. Instead, its presence was represented via appropriate contact interaction definitions, particularly through the assignment of frictional properties (Table 1). To maintain computational efficiency while preserving the relevant contact behaviour, a depth of 10 mm was modelled for the support surface. The constitutive models and mechanical properties of all biological tissues used in the heel simulations were identical to those described in our previous published works [1, 2, 10], and are specified in Table 2 for completeness.

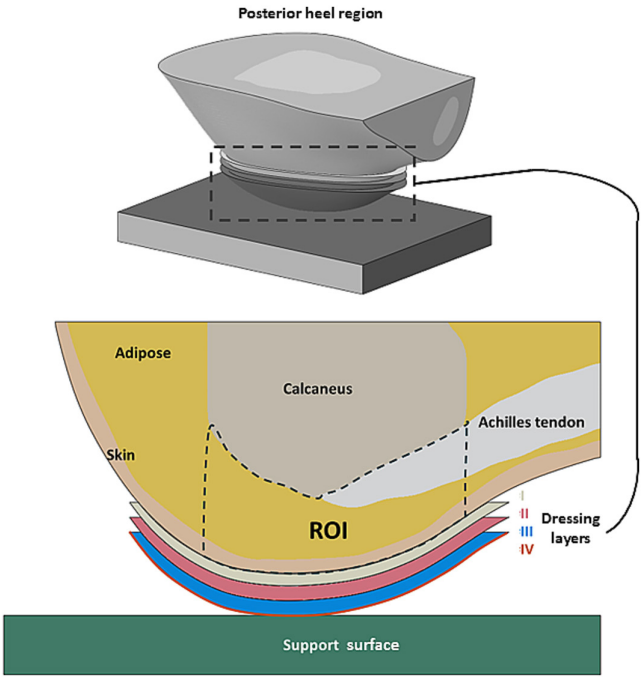

FIGURE 1 | The computational model of the posterior heel region with the applied dressing. The region of interest (ROI) used for evaluating the soft tissue strain and stress distributions is outlined by a dashed black contour.

To characterise the compressive mechanical behaviour of the dressing components, unconfined uniaxial compression tests were conducted on each of the layers I, II and III (separately for each layer) using a modified ASTM D3574 protocol. All tests were performed by means of an electromechanical testing system (Instron 5944, Instron Corp., Norwood, MA, USA) at a quasi-static displacement rate of 12 ± 3 mm/min in a ramp-and-hold loading mode. Specimens were compressed to 50% strain, and this deformation level was held for 300 s to capture the viscoelastic behaviour of the tested material, following which the instantaneous E_{ins} and long-term E_{lt} compressive moduli were derived from the stress–strain and stress-relaxation curves, respectively. The AHT layer was further evaluated under moist conditions: 5%, 10%, and 15% isotonic saline weight gain to simulate perspiration-induced fluid uptake. The dry AHT E_{lt} modulus was 4.1 ± 0.4 kPa, while that of the moist specimens was approximately three-quarters of this value. However, statistical analysis, namely, analysis of variance (ANOVA) and Tukey–Kramer post hoc test, detected no significant differences in E_{lt} across the 5%, 10% and 15% moisture levels, supporting the use of an average modulus of 3.1 ± 0.2 kPa (mean ± standard deviation of the pooled E_{lt} data from the moist specimens) to represent a moist condition in the modelling. The mechanical properties of all the model components are detailed in Table 1.

2.2 | Boundary Conditions, Contact Interactions and the Numerical Method

Boundary conditions were defined to simulate a supine patient resting on a support surface, followed by sliding or

TABLE 1 | Contact conditions between adjacent components in the computational model.

Dressing condition	Contact pair	Contact type	COF
Dry (new)	Skin and Layer I	Frictional	1.55[a]
	Layer I and Layer II	Tied	—
	Layer II and Layer III	Tied	—
	Layer IV and Support surface	Frictional	0.33[b]
Moist (used)	Skin and Layer I	Frictional	0.22[a]
	Layer I and Layer II	Tied	—
	Layer II and Layer III	Tied	—
	Layer IV and Support surface	Frictional	0.33

Note: Layers are the dressing layers as numbered in Figure 1.
Abbreviation: COF = coefficient of friction.
[a]Reported in Gefen et al. (2025).
[b]Experimentally measured in this study using a tilting table tribometer (standard deviation ±0.03).

repositioning in bed, such as when the head of the bed is elevated, or if the patient is repositioned by the nursing staff or moves spontaneously [1, 19]. To simulate such interactions between the posterior heel and the support surface, a vertical displacement of 4 mm was applied to the superior surface of the calcaneal bone, inducing soft tissue compression, followed by 3.5 mm horizontal displacement of the calcaneus to add shear loading (Figure 2, left column). The posterior aspect of the support surface was fully constrained for both translation and rotation movements to simulate the effect of the bedframe. Tied contact constraints were applied between all internal tissue components, representing anatomical attachments to prevent non-physiological tissue motion. In the model variant with the dressing, inter-layer dressing interfaces were also defined as tied. Frictional contact was defined at the interfaces of the dressing with skin and the support surface, respectively. The COF between the external dressing surface (layer IV) and a cotton sheet was experimentally measured using a tribometer and found to be 0.33 ± 0.03 (Table 1). The COF between the skin and the AHT layer was set according to a prior study: 1.55 for the dry (new) dressing condition and 0.22 for the moist (used) condition,[2] representing approximately 5 days of dressing usage (equivalent to a moisture level of at least 10% in the dressing) [30].

TABLE 2 | Mechanical properties of the components of the computational model.

Component	Material type	Elastic modulus E [kPa]	Poisson's ratio ν	Material parameter C_{10}[kPa]	Compressibility parameter D_1 [kPa^{-1}]
Calcaneus bone	Elastic	7×10^6	0.3	—	—
Achilles tendon	Elastic	205	0.49	—	—
Adipose tissue	Hyperelastic	—	—	0.14	7.0×10^{-2}
Skin	Hyperelastic	—	—	15.95	6.3×10^{-4}
AHT dry	Elastic	4.1	0.45	—	—
AHT moist		3.1	0.45	—	—
Superabsorbent layer	Elastic	8.0	0.45	—	—
Foam layer	Elastic	12.0	0.45	—	—
Support surface	Elastic	45	0.3	—	—

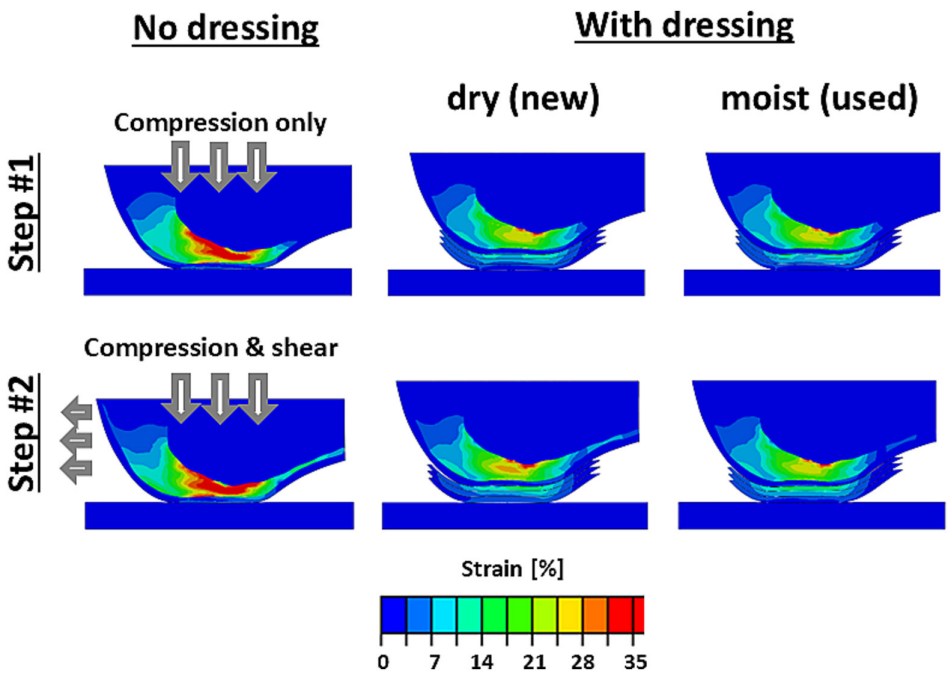

FIGURE 2 | Effective strain distributions in a sagittal cross-section of the posterior heel for the three model variants: No dressing (left column), the tested dressing in its dry (new) state (middle column), and the dressing in its moist (used) state (right column). The simulations were performed in two sequential steps: (i) compressive loading (top row), and (ii) added shear loading (bottom row), the latter representing patient movements such as sliding in bed due to head-of-bed elevation or repositioning.

TABLE 3 | The types and numbers of nodes and elements used for the meshing of the heel tissues, the studied dressing and the support surface.

Model component	Element type	Number of nodes	Number of elements	Degrees of freedom
Calcaneus bone	C3D4H	8346	34446	59484
Achilles tendon	C3D4H	4059	13869	26046
Adipose tissue	C3D4H	69158	282267	489741
Skin	C3D4H	90476	324385	595813
AHT	C3D4H	18059	75811	54177
Superabsorbent layer	C3D4H	20204	89146	60612
Foam layer	C3D4H	20204	89146	60612
Support surface	C3D8R	156996	143000	470988

The meshing technique for the heel tissues was described in earlier publications [1, 19]. The types and numbers of elements used for the meshing of both heel tissues and the dressing are listed in Table 3. All the simulations were carried out using the standard static FE solver in Abaqus/CAE 2023, which is suitable for solving nonlinear, large-deformation problems. Each model variant required approximately 2 h of computation on a 64-bit Windows 10 workstation equipped with an Intel Core i9-7900X processor (3.30 GHz) and 64 GB of RAM, which is typical for the level of computational complexity of this study.

2.3 | Outcome Measures

We evaluated the volumetric exposure of the soft tissues in the posterior heel to a sustained mechanical loading, with specific focus on the Green-Lagrange strain and Cauchy stress distributions. The analysis was performed for skin, adipose and the Achilles tendon tissues within a predefined region of interest (ROI) (Figure 1). Histogram plots were generated to represent the distributions of soft tissue strain and stress magnitudes within each tissue type, as well as for the combined soft tissues of the heel, i.e., skin, adipose and the Achilles tendon within the ROI taken together.

To quantify the protective performance of the studied dressing, we computed its protective efficacy index (PEI), a metric originally developed by the lead author and widely used in the wound care literature for evaluating PUs prevention technologies, including prophylactic dressings and positioners [16]. The PEI metric is a dimensionless biomechanical performance measure that quantifies the percentage reduction in soft tissue strain or stress exposure achieved by a dressing relative to a no-dressing condition, with higher PEI values indicating greater protective efficacy. The PEI therefore reflects how effectively a specific dressing design alleviates soft tissue strain and stress exposure compared with an equivalent no-dressing scenario. To calculate the PEI data for the dressing under investigation, we first computed the areas under the strain and stress histograms for the no-dressing and dressing model variants (Figures 3). The percentage reduction in area under the histogram curve, indicating the degree of histogram flattening due to the presence of the dressing on the heel, was then used to quantify the protective capacity. A more pronounced histogram flattening indicates a higher PEI and thus greater protective benefit. The PEI values were calculated separately for two domains (Figures 4): (i) the upper three quartiles (25th to the 100th-percentile) of the tissue strain and stress domains, to exclude low-level tissue loading, and (ii) for the top quartile (75th to the 100th-percentile) of the strain and stress domains, to isolate and assess the protective effect of the dressing on the highest strain and stress concentrations.

3 | Results

The effective strain distributions in the heel soft tissues (Figure 2) revealed strain concentrations beneath the calcaneus, which intensified when shear was applied. However, with applied dressing, remarkable alleviation of these concentrated strains occurs, particularly when the dressing is simulated to be moist. Histograms of soft tissue exposure within the ROI quantitatively indicate that the application of the studied dressing substantially reduced strain and stress exposures above the 25th-percentile, and in both skin and adipose tissues (Figure 3). This effect of tissue loading alleviation was observed in both the dry (new) and moist (used) simulated dressing conditions but was more pronounced in the moist (used) state (Figures 3 and 4).

The PEI for the strain- and stress-based calculations above the 25th-percentile ranged from 71.8% to 88.7% and from 73.4% to 99.9%, respectively, depending on the specific soft tissue type and dressing condition (Table 4). Notably, in the loading subdomains reflecting the greatest tissue injury risk, i.e., above the 75th-percentile of the strain and stress domains, the dressing was associated with an absence of high-magnitude strain and stress concentrations within the analysed loading domain under the simulated conditions (Table 4 and Figure 4). Importantly, according to all the PEI measures, the dressing delivered increased protective efficacy when simulated to be moist (used) compared to when it was dry (new) (Table 4 and Figure 4).

To assess how the soft tissue loading evolves under shear, we plotted the average maximal principal strain and stress data in skin and adipose tissues within the ROI as a function of the imposed shear displacement and fitted regression lines (Figure 5). For the no-dressing condition, both strain

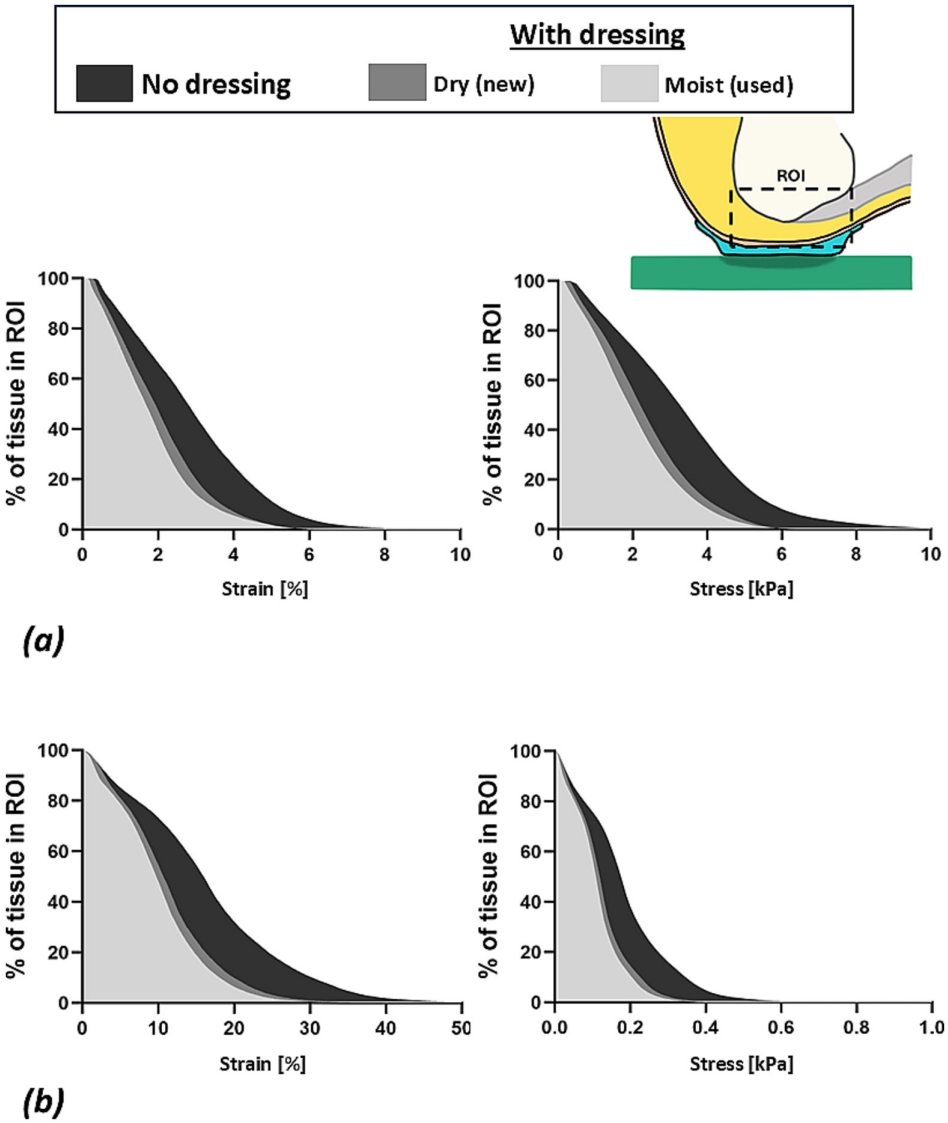

FIGURE 3 | Histograms showing exposure of (a) skin and (b) adipose tissue to the full range of effective strains (left column) and stresses (right column) in the region of interest (ROI), comparing the no-dressing condition to conditions where the tested dressing was applied prophylactically at its dry (new) and moist (used) conditions.

and stress increased monotonously with shear, particularly in the skin. With the dressing in its moist (used) condition, the regression lines remained nearly flat, indicating that increasing shear had a negligible effect on tissue loading. These findings suggest that the dressing attenuates soft tissue strain and stress build-up during frictional loading from body movements to a greater extent under moist conditions than under dry conditions.

4 | Discussion

This study demonstrated computationally that the multilayer prophylactic dressing investigated here can meaningfully reduce modelled mechanical risk factors for heel PU formation, particularly those associated with deep tissue injury [33–37]. We specifically demonstrated that the studied dressing substantially mitigates peak soft tissue loads in the posterior heel under conditions representative of clinical use. The FE simulations revealed pronounced alleviation of both strain and stress magnitudes throughout the heel soft tissues when the dressing was applied, compared with the no-dressing condition. The histograms of tissue exposure indicated considerable reductions above the 25th-percentile of both soft tissue strain and stress distributions, with marked attenuation of high-magnitude concentrations above the 75th-percentile, including in subdermal tissues, so that no such concentrations were observed within the upper-quartile loading range under the modelled conditions (Figure 4). These findings support the biomechanical capacity of the studied dressing to protect at-risk heel tissues from PUs and in particular, from the deep tissue injury type which is associated with focal, elevated loading in deep soft tissues [38]. As with other multilayered dressings, the pronounced protective performance can be partially attributed to its composite architecture containing materials with contrasting stiffness properties: the compliant, absorptive, and cushioning layers with varied stiffness properties work synergistically to redistribute and dissipate applied mechanical

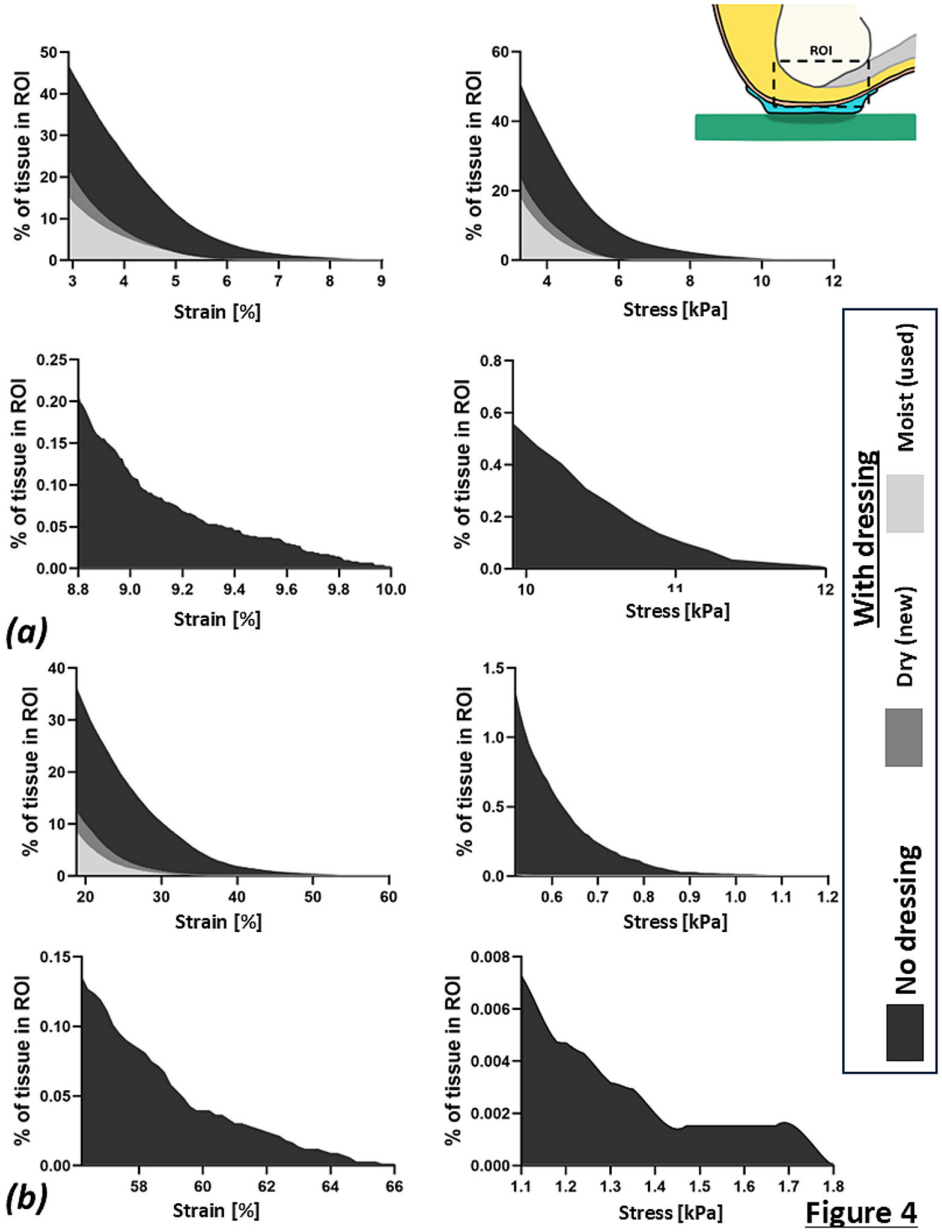

FIGURE 4 | Histograms of soft tissue exposure to loading, for (a) skin and (b) adipose tissue exposed to effective strains (left column) and stresses (right column) within the region of interest (ROI). For each panel, the top and bottom rows represent exposure to soft tissue strain/stress values exceeding the 25th-percentile and the 75th-percentile, respectively, for the no-dressing condition compared to conditions where the tested dressing was applied prophylactically at its dry (new) and moist (used) conditions.

loads [1, 16, 19]. Nevertheless, the magnitude of the protective effect indicated above, particularly under moist conditions, should be mainly attributed to the contribution of the moisture-responsive frictional adaptation in the AHT layer. Specifically, the observed protective effect under moist conditions is primarily due to the reduced COF of the AHT layer when hydrated (Table 4). Under moist (used) conditions, the PEI values were consistently higher for both the strain- and stress-based calculations across all the soft tissues analysed (Table 4), which coincided with the ~8-fold reduction in the COF between the skin and AHT layer from dry to moist conditions (Table 1). The lower COF of the moist AHT with skin facilitates controlled micro-sliding at the skin-dressing interface, thereby reducing the transfer of frictional forces to the skin and subcutaneous tissues and hence the formation of shear stresses. In other words, this COF reduction mechanism transforms the moist skin-contacting layer in the dressing, micro-gelling into a self-lubricating interface that internally dissipates frictional loads rather than transmitting them to the skin and subcutaneously. These results agree with previously published work, indicating that a COF of skin-interacting materials below 0.4 reduces the maximum shear strains in the soft tissues of the posterior heel, and that higher COFs effectively lead to a near-no-slip condition [39]. The current findings suggest that dynamic frictional adaptability to moisture may contribute to maintaining biomechanical protection during prolonged wear [30].

The ability of the studied dressing to decouple external frictional forces from internal soft tissue deformations was further demonstrated in the shear displacement analysis (Figure 5).

TABLE 4 | The protective efficacy index (PEI) of the tested dressing under dry (new) and moist (used) conditions.

	PEI [%]			
	Above 25%		Above 75%	
	Dry (new)	Moist (used)	Dry (new)	Moist (used)
Strain-based calculation				
Skin	71.8	78.2	All the *strain-based* PEI values reached 100% within the upper quartile loading domain as defined by the present metric	
Adipose tissue	81.3	88.7		
Heel soft tissues combined	80.9	88.5		
Stress-based calculation				
Skin	73.4	81.0	All the *stress-based* PEI values reached 100% within the upper quartile loading domain as defined by the present metric	
Adipose tissue	99.3	99.9		
Heel soft tissues combined	71.8	80.4		

Under moist conditions, tissue strain and stress responses remained nearly constant, indicating effective shear damping by the dressing, whereas in the no-dressing condition, both tissue strains and stresses increased progressively with the shear displacement (Figure 5). These findings are consistent with the moisture-responsive friction adaptation mechanism described above, whereby reduced interfacial friction limits shear transfer to underlying tissues. This shear-damping behaviour represents a biomechanically advantageous mechanism that stabilises the local mechanical environment under conditions mimicking patient repositioning, especially during sub-optimal repositioning (where the patient is pulled along the bed; a common issue in patients with impaired mobility), or while a patient is sliding in their bed when the head of the bed is elevated. Such movements are well established key risk factors in heel PU development [35, 39–42].

While the present FE modelling framework enabled detailed, tissue-level analysis of the biomechanical function of the studied dressing in protecting from heel PUs, several limitations exist. Importantly, FE modelling provides mechanistic predictions under defined assumptions and boundary conditions rather than direct evidence of clinical effectiveness and, accordingly, translation of these findings to clinical outcomes requires further research, including studies in relevant patient subpopulations.

Firstly, the moisture effects were modelled using average COF values derived from tribometry measurements, and these may not fully capture temporal changes in dressing hydration due to perspiration responses of individual patients. The moist condition considered here was intended to represent mild perspiration-associated interface hydration during prolonged wear rather than persistently moist heel skin. For example, patients with neural damage resulting from diabetes, spinal cord injury or other peripheral or central nerve disorders, as well as with skin damage from burns or radiation, may not sweat or have reduced sweating, which may not reduce the COF of the AHT with their skin to the level assumed here. Likewise, patients with vascular disease, autonomic dysfunction, neuropathy, or altered skin physiology may exhibit different hydration responses at the skin-dressing interface, potentially influencing interfacial friction behaviour differently from that assumed in the present simulations.

Second, the current modelling work did not consider exposure of the dressing to non-sweat body fluids, e.g., urine or exudate from a nearby wound, which may alter the COF values differently. However, unlike sacral or perineal anatomical regions, the heel is generally less exposed to moisture sources such as urine or stool and, therefore, perspiration-related moisture was considered the most clinically relevant source of interface hydration for this anatomical site. In this context, experimental overnight measurements reported only minimal dressing mass gain (~0.3 g) [20], consistent with limited interface hydration during wear. Accordingly, the moist condition modelled in this study should be interpreted as reflecting limited perspiration-related interface hydration during prolonged wear at the heel, rather than persistently moist skin or exposure to alternative fluid sources such as wound exudate or incontinence-associated moisture.

Third, patient-specific anatomical and physiological variability, including foot deformities, edema, contractures, scars from previous wounds, vascular compromise, neuropathy, or other conditions affecting heel anatomy, morphology, tissue composition, mechanical properties or loading patterns, as well as uncharacteristic movement behaviours (e.g., due to psychiatric agitation), were not accounted for. Accordingly, the present FE model, which was based on a standard anatomical configuration under controlled loading conditions, may not fully capture the range of patient presentations encountered clinically.

Fourth, inadequate support surface conditions such as mattress sagging caused by wear-and-tear or a hammock effect of the bedsheet could influence the performance of the studied dressing; these external influences were beyond the scope of the present, controlled FE modelling conditions. Furthermore, the present modelling approach assumed static hydration conditions and therefore did not account for patient-specific temporal changes in moisture accumulation or hydration behaviour over prolonged wear periods, nor for factors that may influence

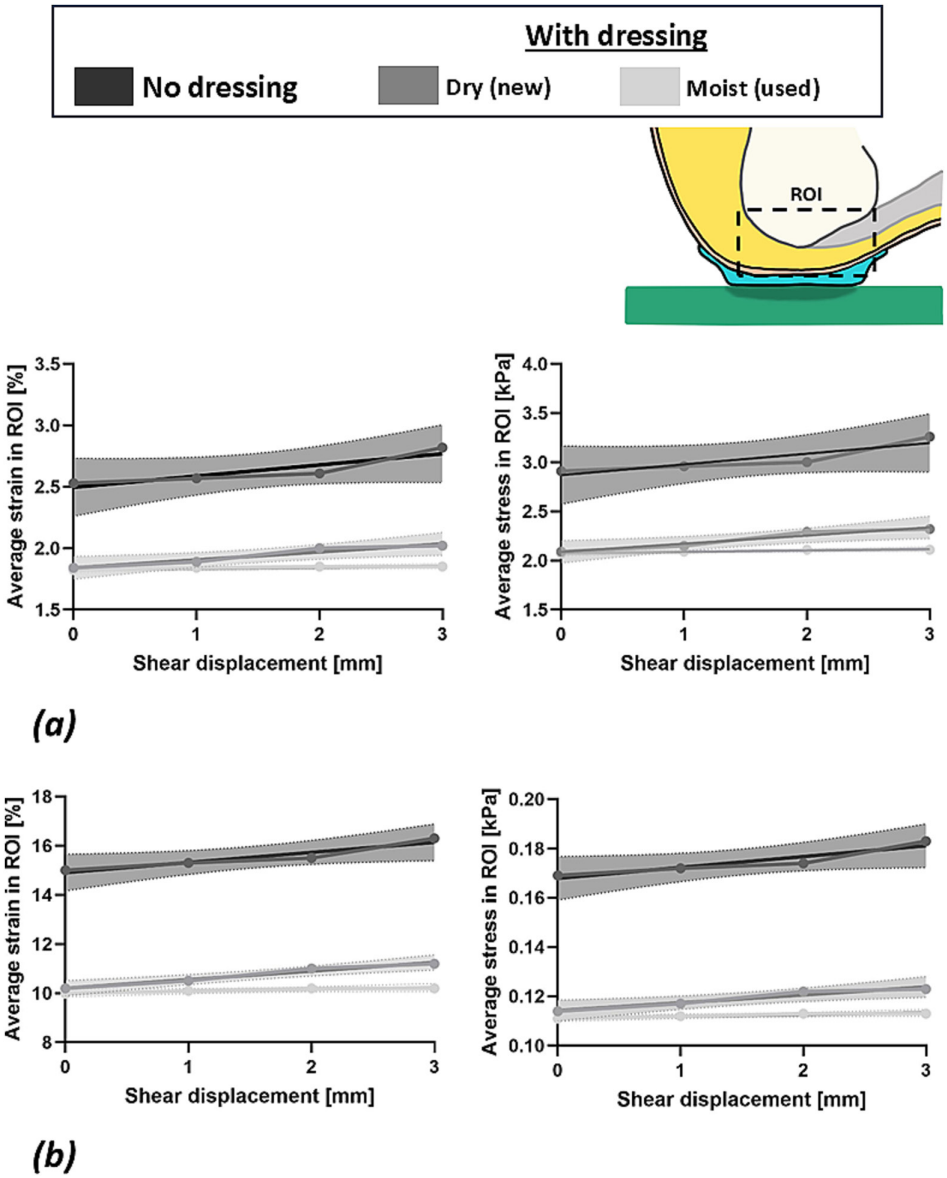

FIGURE 5 | Plots of the average maximal principal strain (left column) and stress (right column) in (a) skin and (b) adipose tissue within the region of interest (ROI) as a function of the shear displacements, comparing the no-dressing condition to conditions where the tested dressing was applied prophylactically at its dry (new) and moist (used) conditions.

local interface hydration, such as gravity-driven moisture redistribution or support surface geometry. Future modelling efforts could therefore integrate dynamic hydration behaviours, patient-specific sweat rates, and evolving moisture exposure scenarios, as well as interactions between the heel anatomy, tissue composition and biomechanics and the body posture, local moisture redistribution within the dressing, and the support surface geometry and deformation, to improve the clinical relevance of such models.

Lastly, no direct comparator prophylactic dressing with a silicone-based skin-contact interface was modelled. Therefore, the findings should not be interpreted as evidence of superiority or inferiority relative to dressings incorporating silicone-containing skin-contact layers, nor as a basis for informing modifications to current guideline recommendations. Rather, the present work examines the biomechanical implications of moisture-responsive friction adaptation within a single dressing design. Future studies incorporating direct comparisons between dressing technologies and products with differing interfacial friction characteristics are warranted. Such comparative studies would also help determine whether differences in interfacial friction properties translate into meaningful differences in soft tissue protection and, ultimately, clinical effectiveness and cost-effectiveness outcomes.

5 | Conclusions

This study presents in silico evidence, supported by experimental laboratory data, that the dressing under investigation considerably reduces strains and stresses in heel soft tissues under simulated loading conditions representative of typical supine positioning and patient migration in bed, with greater biomechanical protective efficacy under moist conditions. The current findings also suggest a potential role for moisture-adaptive

interfacial materials (also termed 'smart materials') in next-generation prophylactic dressings aimed at preventing PUs in at-risk populations [43, 44]. Further studies are warranted to determine whether the modelled biomechanical effects observed under varying interfacial hydration conditions translate into clinically meaningful outcomes in relevant patient populations. Nonetheless, the present work did not assess clinical effectiveness, cost-effectiveness, or the broader justification for prophylactic use of advanced wound care technologies on intact skin. Rather, the study focused on mechanistic evaluation of dressing-skin interface behaviour under conditions relevant to prophylactic applications. Whether specific dressing technologies provide clinically meaningful incremental benefits relative to their design complexity, overall resource utilisation, or implementation burden therefore remains an important topic for future clinical and health-economic research.

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Ethics Statement

The authors have nothing to report.

Consent

The authors have nothing to report.

Conflicts of Interest

Author A.G. is a paid consultant of Convatec Ltd. The other author declares 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

[1] Certain clinical priorities require head-of-bed elevations, such as for preventing aspiration pneumonia and managing gastroesophageal reflux, and these typically overweigh the need to prevent patient migration in bed.

[2] The measured static and kinetic coefficients of friction (COF) were $\mu_s = 0.22 \pm 0.06$ and $\mu_k = 0.19 \pm 0.02$, respectively [30]. As the contact defined in the finite element modelling between the dressing and support surface is continuously sliding, the kinetic coefficient, μ_k, governs the response. Considering the experimental uncertainty (at least ± 0.02) and the overlap between the static and kinetic ranges (0.17–0.21), a nominal COF value of 0.2 was adopted. This value lies within the measured variability and avoids implying unwarranted precision in a parameter that is inherently uncertain in contact (bio)mechanics and in the context of expected support surface variabilities.

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