Basic research

The biomechanical efficacy of dressings in preventing heel ulcers

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KEYWORDS
Heel pressure ulcers; Deep tissue injury; Prophylactic dressings; Computational modeling; Finite element analysis

Abstract
The heels are the most common site for facility-acquired pressure ulcers (PUs), and are also the most susceptible location for deep tissue injuries. The use of multilayer prophylactic dressings to prevent heel PUs is a relatively new prevention concept, generally aimed at minimizing the risk for heel ulcers (HUs) through mechanical cushioning and reduction of friction at the dressing-support interface. We used 9 finite element model variants of the posterior heel in order to evaluate the biomechanical performance of a multilayer dressing in prevention of HUs during supine lying. We compared volumetric exposures of the loaded soft tissues to effective and maximal shear strains, as well as peak stresses in the Achilles tendon, without any dressing and with a single-layer or a multilayer dressing (Mepilex® Border Heel-type), on supports with different stiffnesses. The use of the multilayer dressing consistently and considerably reduced soft tissue exposures to elevated strains at the posterior heel, on all of the tested support surfaces and when loaded with either pure compression or combined compression and shear. The aforementioned multilayer design showed (i) clear benefit over a single-layer dressing in terms of dissipating tissue strains, by promoting internal shear in the dressing which diverts loads from tissues; (ii) a protective effect that was consistent on supports with different stiffnesses. Recent randomized controlled trials confirmed the efficacy of the simulated multilayer dressing, and so, taken together with this modeling work, the use of a prophylactic multilayer dressing indicates a great promise in taking this route for prevention.

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1. Introduction
Pressure ulcers (PUs) are localized injuries resulting primarily from sustained soft tissue deformations [1]. Heel ulcers (HUs) are the most
common facility-acquired PUs; the heel is also where the prevalence of deep tissue injuries (DTIs) is the greatest [2–5]. In a supine posture, skin and subdermal fat near the posterior aspect of the calcaneus bone are generally subjected to sustained compressive and shear loads due to the weight of the foot and horizontal reaction forces, e.g. due to elevation of the head of the bed. The relatively thin and hence highly deformed soft tissue between the calcaneus and support are then at risk for HUs, particularly to DTI [6]. Populations at-risk are the elderly, individuals with neurological impairments and/or restricted mobility, as well as surgical patients, as these individuals are more likely to spend long periods in a static supine position [7].

Finite element (FE) modeling is a powerful tool which allows to evaluate internal tissue loads and isolate the influence of biomechanical characteristics from other potential risk factors e.g. impaired circulation or tissue repair capacities [8–12]. In a previous relevant FE modeling work published by our group, Sopher and colleagues [12] studied the effects of foot posture and support stiffnesses on strains and stresses within fat tissues of a resting heel; however, they did not include the Achilles tendon (AT), which may substantially influence the mechanical state of nearby soft tissues. Tenenbaum et al. [13] later used MRI to show that technologies and devices for protecting the heel against HUs differ in extents of soft tissue deformations that they induce, but the analysis was limited to heel boots.

Since PUs are difficult and costly to treat, efforts are focused on developing effective prevention strategies [14,15]. While the consensus is that complete off-loading of the heels is the most effective method for preventing HUs [6], off-loading is not always feasible, especially with moving patients. The use of prophylactic dressings to protect the heel against HUs is a relatively new prevention concept, which is focusing growing attention from different research groups [6,16,17]. Several studies have recently demonstrated the effectiveness of dressings in the prevention of HUs by means of randomized controlled trials (RCTs) and laboratory measurements of contact pressure and shear in experimental models [18–21]. However, no study yet has investigated states of mechanical loading at the soft tissues of the supported heel, and how these are influenced by dressings. Three-dimensional (3D) MRI-based FE modeling was proven efficient for such studies; indeed, this has been the goal of the present work.

2. Methods

2.1. Geometry

In order to evaluate the biomechanical performance of a multilayer heel dressing for the prevention of HUs, a set of 9 FE model variants was developed (Table 1). Each of the model variants included the posterior part of the calcaneus bone, the AT, the fat and the skin of the left heel, as well as a flat foam support (Fig. 1a,b). Model variants #1–4 did not include any dressing and were used as ‘reference’ cases for the purpose of comparisons to cases where dressings were present (Table 1). Variant #5 included a single-layer foam dressing and variants #6–9 included a multilayer dressing enveloping the posterior aspect of the heel (Table 1, Fig. 1b).

Fifty-six T1-weighted axial MRI slices from the suspended left heel of a healthy male subject (subject #2 in the Tenenbaum et al. [13] study: age 34 years; weight 90 kg) were used to develop the anatomically-realistic 3D model geometry. Additional information regarding the MRI system and scan protocol is available in our previously published work [13]. Here, we used the ScanIP® module of Simpleware® [22] to segment the different tissues from the MRI dataset, and then, to truncate the model volume in order to focus on the retro-calcaneal region of the foot (Fig. 1a,b). In model variants #6–9 we represented a multilayer dressing. To make these simulations more realistic and clinically relevant, we modeled the 5-layer Mepilex® Border Heel dressing (Mölndlycke Health Care, Gothenburg, Sweden), which is one of the few dressing products now being marketed with intention to prevent HUs,1 in addition to treatment. The innermost and outermost layers in this particular product, i.e. the ‘backing film’ (layer I; Fig. 1c) which faces the support and ‘Safetac®’ layer which attaches to the skin (layer V), are approximately 20-fold thinner than the internal 3 layers. Hence, these external two layers were modeled as contact conditions (see Section 2.3) rather than physical layers. The thicknesses of the 3 internal layers (taken as physical layers for the purpose of the modeling) were set according to the manufacturer’s specifications, as follows: Airlaid = 2.2 mm (layer II; Fig. 1c), non-woven layer = 0.2 mm (layer III), polyurethane (PUR) foam = 1.6 mm (layer IV). In order to determine if a multilayer dressing offers additional value in reducing internal tissue loads with respect to a

1 In conjunction with regular prevention protocols.
single-layer foam dressing, we used model variant #5, where we incorporated a single-layer dressing with thickness that is the total of the above 3 physical layers. The geometry of the flat support (thickness 10 mm) was added in all the model variants at the pre-processing stage, in PreView of FEBio [23].

2.2. Mechanical properties

Constitutive laws and mechanical properties of all tissues were adopted from the literature (Table 2). Specifically, the calcaneus and AT were assumed to be linear-elastic isotropic materials with elastic moduli of 7 GPa and 205 kPa, and Poisson’s ratios of 0.3 and 0.49, respectively [24,25] (Table 2). Skin and fat tissues were assumed to be nearly-incompressible (Poisson’s ratio of 0.495), nonlinear isotropic materials with their large deformation behavior described using an uncoupled Neo-Hookean material model [26] with a strain energy density function $W$:

<table>
<thead>
<tr>
<th>Model variant</th>
<th>Dressing</th>
<th>Elastic modulus of support</th>
<th>Loading mode</th>
</tr>
</thead>
<tbody>
<tr>
<td>#1</td>
<td>Off</td>
<td>45 kPa</td>
<td>Compression</td>
</tr>
<tr>
<td>#2</td>
<td>Off</td>
<td>63 kPa</td>
<td>Compression</td>
</tr>
<tr>
<td>#3</td>
<td>Off</td>
<td>80 kPa</td>
<td>Compression</td>
</tr>
<tr>
<td>#4</td>
<td>Off</td>
<td>63 kPa</td>
<td>Compression + Shear</td>
</tr>
<tr>
<td>#5</td>
<td>Single-layer</td>
<td>63 kPa</td>
<td>Compression</td>
</tr>
<tr>
<td>#6</td>
<td>Multilayer</td>
<td>45 kPa</td>
<td>Compression</td>
</tr>
<tr>
<td>#7</td>
<td>Multilayer</td>
<td>63 kPa</td>
<td>Compression</td>
</tr>
<tr>
<td>#8</td>
<td>Multilayer</td>
<td>80 kPa</td>
<td>Compression</td>
</tr>
<tr>
<td>#9</td>
<td>Multilayer</td>
<td>63 kPa</td>
<td>Compression + Shear</td>
</tr>
</tbody>
</table>

Table 1 The model variants.

Fig. 1 Finite element computational modeling of the left heel with the Mepilex® Border Heel (Möllycke Health Care, Gothenburg, Sweden) multilayer dressing: (a) An axial MRI slice from the three-dimensional (3D) scan set which has been used for reconstructing the model geometry. The modeled volume is marked by a dashed frame. (b) A sagittal cut through the 3D model of the heel and aforementioned multilayer dressing. CB = calcaneus bone, AT = Achilles tendon, F = fat, S = skin. The volume of interest (VOI) is indicated using a dashed frame. (c) The tetrahedral mesh of the heel and multilayer dressing (right panel) and a zoom-in on the structure of the dressing (left panel), identifying the individual layers which were represented as follows: I = frictional sliding contact with the support, II–IV = physical elastic layers of the dressing (with properties as detailed in Table 2), V = tied interface with the skin.

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\[ w = \frac{G_{ins}}{2} \left( \lambda_1^2 + \lambda_2^2 + \lambda_3^2 - 3 \right) + \frac{1}{2} K (\ln J)^2 \]  

where \( G_{ins} \) is the instantaneous shear modulus (Table 2), \( \lambda_i \) (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 2) [24,27]. The flat foam support was assumed to be isotropic linear-elastic with a Poisson’s ratio of 0.3 and elastic moduli of 45 kPa, 63 kPa or 80 kPa [28].

We measured the individual stiffnesses of each physical layer of the multilayer dressing (II, III, IV; Fig. 1c) in unconfined compression, using an electromechanical uniaxial testing system (INSTRON Co. model 5544; High Wycombe, UK) at a quasi-static strain rate of 20%/min. Corresponding elastic moduli were calculated by linearly approximating the stress–strain curves for the 20%–50% strain domain, representing the expected (large deformation) use in clinical settings [13] (Table 1). The elastic modulus of the single-layer foam dressing was set as the weighted average of the elastic moduli of the multilayer dressing components II, III, IV with their respective thicknesses to allow comparisons. Poisson’s ratios of all dressing materials were set as 0.258 [29].

### 2.3. Boundary and material transition conditions

Boundary conditions were chosen to simulate the descent/shear of the weight-bearing (WB) foot during supine lying, on flat foam supports. For calibration, we used the difference in total soft tissue thickness (skin, fat and tendon) between non-WB and WB MRI datasets, as the target downward displacement for the calcaneus against a rigid support. We further verified agreement between the resulting thickness of fat tissue alone in our WB simulation against fat thickness in the WB MRI, and recorded the heel-support reaction force. Since this reaction force equals, in practice, to the weight of the foot/ankle complex in the supine position, we used downward displacements in the range of 4.4–5.1 mm in the model variants to produce the same heel-support reaction forces on all the elastic (non-rigid) supports (Fig. 2a). In variants #4 and #9, we also included horizontal displacements imposed on the superior surface of the calcaneus that were arbitrarily chosen to be of the same magnitude as the vertical displacement, in order to represent shearing forces acting on the foot when the head of the bed is elevated or during repositioning (Table 1). The bottom surface of the elastic supports was fixed for all translations and rotations. Tied interfaces were defined between all tissue components. The coefficient of friction (COF) between bare skin and the support was set as 0.43 to mimic typical dynamic skin/cotton-sheet friction, as the final strains and stresses in the soft tissues of the resting heel in our simulations form at the end of a positioning/repositioning maneuver which lasts for a short time [30]. The backing film (layer I) of the multilayer dressing, which is designed to reduce the skin-support friction, was represented as frictional sliding contact, hence the COF between physical layer II and the support was set as 0.35 (Figs. 1c and 2a) [29].

The Safetac® layer (layer V) of the multilayer dressing, which is attached to the skin, was modeled as a tied

### Table 2  Mechanical properties of the model components and characteristics of the finite element mesh.

<table>
<thead>
<tr>
<th>Modal component</th>
<th>Shear modulus [kPa]</th>
<th>Bulk modulus [kPa]</th>
<th>Elastic modulus [kPa]</th>
<th>Poisson’s ratio</th>
<th>Numbers of mesh elements</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skin</td>
<td>31.9</td>
<td>3179.37</td>
<td>—</td>
<td>0.495</td>
<td>41,503–1,27,240</td>
</tr>
<tr>
<td>Fat</td>
<td>0.286</td>
<td>28.5</td>
<td>—</td>
<td>0.495</td>
<td>81,208–99,953</td>
</tr>
<tr>
<td>Tendon</td>
<td>—</td>
<td>—</td>
<td>205</td>
<td>0.49</td>
<td>12,123–13,159</td>
</tr>
<tr>
<td>Bone</td>
<td>—</td>
<td>7 \times 10^6</td>
<td>0.3</td>
<td></td>
<td>25,979–26,583</td>
</tr>
<tr>
<td>Airlaid (layer II)</td>
<td>—</td>
<td>—</td>
<td>15.3</td>
<td>0.258</td>
<td>173,407</td>
</tr>
<tr>
<td>Non-woven (layer III)</td>
<td>—</td>
<td>—</td>
<td>75</td>
<td>0.258</td>
<td>123,448</td>
</tr>
<tr>
<td>Polyurethane foam (layer IV)</td>
<td>—</td>
<td>—</td>
<td>12</td>
<td>0.258</td>
<td>214,710</td>
</tr>
<tr>
<td>Single-layer foam</td>
<td>—</td>
<td>—</td>
<td>19</td>
<td>0.3</td>
<td>16,554</td>
</tr>
<tr>
<td>Support</td>
<td>—</td>
<td>45</td>
<td>45</td>
<td>0.3</td>
<td>121,500</td>
</tr>
</tbody>
</table>

2 We used the average value of the dynamic COF that was measured between two Mepilex® Border Sacrum dressings, as they both have the same outermost layer as that of the Mepilex® Border Heel.

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interface condition between physical layer IV of the dressing and the skin.

2.4. Numerical method

Meshing of the tissues and dressings was performed using the ScanIP\textsuperscript{®} module of Simpleware\textsuperscript{®} [22], with finer meshes used in the skin and dressing materials around the contact area of the dressing, or bare skin, with the support (Fig. 1c). Meshing of the supports was performed in the Preview module of FEBio [23]. The runtime of each model variant was 8–34 h using a 64-bit Windows 8-based workstation with 2 × Intel Xeon E5-2620 2.00 GHz CPU and 64 GB of RAM.

2.5. Outcome measures

We compared volumetric exposures of the soft tissues to effective Green-Lagrange strains and maximal shear strains, with the effective strain defined as [26]:

\[
E_{\text{eff}} = \sqrt{E_{xx}^2 + E_{yy}^2 + E_{zz}^2 - E_{xx}E_{yy} - E_{yy}E_{zz} - E_{zz}E_{xx} + 3 \left( E_{yy}^2 + E_{zz}^2 + E_{xx}^2 \right)}
\]  

Fig. 2  (a) The loading conditions applied to the three-dimensional model of the heel with the multilayer dressing. CB = calcaneus bone, F = fat, S = skin, D = dressing. (b) Comparison of compressive strains on the top surfaces of the supports under the heel, for the bare skin cases.
Strain data were pooled from tendon, fat and skin tissues together, for all the elements in a volume of interest (VOI) defined by the circumference of the calcaneus and its projection at the retro-calcaneal region, as depicted in Fig. 1b. Additionally, we compared the peak effective, compressive and shear Cauchy stresses in the AT, the effective stress being defined as [26]:

\[
\sigma_{\text{eff}} = \sqrt{\sigma_{xx}^2 + \sigma_{yy}^2 + \sigma_{zz}^2 - \sigma_{xx}\sigma_{yy} - \sigma_{yy}\sigma_{zz} - \sigma_{zz}\sigma_{xx} + 3\left(\frac{\sigma_{xy}^2 + \sigma_{yz}^2 + \sigma_{zx}^2}{C_0}\right)}
\]

Compressing time steps were chosen for data collection so that the resulting reaction forces were within less than a 4%-difference from the target reaction force. We evaluated the biomechanical efficacy of the dressings as the reduction in volumetric exposures to effective and maximal shear strains in the soft tissues contained in the VOI, as well as in the reduction of peak stresses occurring in the AT.

### 3. Results

Compressive strain distributions on the superior surfaces of the flat foam supports, for the bare skin cases, are shown in Fig. 2b. The average contact pressures under the heel, on the different supports, with or without the multilayer dressing, are detailed in Table 3. These values of contact pressure and resulting support strains fit nicely with the recent empirical pressure-strain curves reported by Li and colleagues [31] for foam cushion materials, which validates the modeling.

A comparison of the distributions of effective and maximal shear strains which develop in the soft tissues of the heel on a 63 kPa support is shown in Fig. 3. Both the effective and maximal shear strains peaked at around the calcaneus-fat interface, across all model variants (Fig. 3). Fat

![Fig. 3](image)

Fig. 3 Comparison of states of mechanical loading in the soft tissues of the heel, with and without the multilayer dressing, on a 63 kPa support. (a) Effective strain distributions when the heel was loaded in pure compression. (b) Maximal shear strain distributions when the heel was loaded in combined compression and shear. The outline of the Achilles tendon is marked by a dashed line.
around the AT was also subjected to elevated strains (Fig. 3). Alleviation of peak/large strains with the use of a multilayer dressing is evident in both pure compression and combined compression and shear loading configurations, with the most pronounced effect obtained in subdermal fat tissue. Moreover, the use of a multilayer dressing consistently and considerably reduced the internal soft tissue exposures to elevated strain levels for all the loading configurations, and on all the supports that were tested (Figs. 4–7).

Although the single-layer foam dressing was able to mildly reduce the volumetric exposure of soft tissues to effective strains, the multilayer dressing had a substantially more profound protective effect (Fig. 5). Hence, the simulations clearly indicate that there is added-value in a multilayer design with respect to a single-layer design.

When loaded by combined compression and shear, the multilayer dressing was again able to considerably reduce the volumetric exposure to maximal shear strains (Fig. 6). The reduction in volumetric exposures to effective strains between 0.2 and 0.5, and, separately, the reduction in exposures to effective strains exceeding 0.5 when a multilayer dressing is used, are shown in Fig. 7. Detailed analyses of the ‘protective ratios’, quantifying the added tissue protection provided by the multilayer dressing with regard to exposure to soft tissue strains above level ‘X’ (with respect to a ‘no dressing’ condition), showed that for effective strains exceeding 0.3, 0.5 and 0.7, the multilayer dressing had protective ratios of approximately 2, 3 and 6.5, respectively. The protective ratios for maximal shear strains exceeding 0.2, 0.3 and 0.4 were approximately 2, 2.5 and 15. Therefore, the multilayer dressing specifically reduces exposures to the high-end domain of strains, and so, if tissue strain levels on a given support rise, the protective efficacy of the dressing will be more substantial.

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Effective stress distributions in the AT on a 63 kPa support are shown in Fig. 8. Overall, stresses peaked near the bone-tendon-fat interface, in the lateral aspect of the Achilles insertion site. The multilayer dressing was able to considerably reduce the peak effective, compressive and maximal shear stresses when the heel was loaded in pure compression, on all the tested supports (Fig. 9). Furthermore, when the heel was loaded by combined compression and shear, the multilayer dressing was again able to reduce the effective, compressive and maximal shear stresses, by 49%, 36% and 48%, respectively.

4. Discussion

In the present study, we used 9 MRI-based FE model variants of the posterior heel in order to evaluate the biomechanical efficacy of heel dressings for prevention of HUs. We chose to model the Mepilex® Border Heel dressing (Mölnlycke Health Care), which is now marketed with indication for prevention, to address a real-world problem. We compared volumetric exposures of soft tissues enveloping the retro-calcaneal region.
to elevated strains during supine lying, when using a single-layer foam dressing, a multilayer (Mepilex® Border Heel) dressing or with no dressing. The questions in mind were (i) whether, in biomechanical terms, a dressing is a valid approach for lowering the exposures to tissue strains, and if so, (ii) whether there is additional value in using multilayer over single-layer dressings.

Overall, the peak strains occurred in the subdermal fat adjacent to the bone-fat interface, consistent with our previous modeling work [12], however, here we improved the heel modeling substantially by incorporating the AT and its insertion site. Including the AT inevitably left a thinner fat tissue layer in the proximal retrocalcaneal region, and so, less cushioning is facilitated by fat tissue in that area. Indeed, the stiffer tendon insertion concentrated fat tissue loads around it (Fig. 3), hence, inclusion of the Achilles is essential for developing HUs-related simulations.

We found that the use of a multilayer heel dressing consistently and substantially reduced the volumetric exposure of the tendon, subdermal fat and skin to elevated strain levels. This implies that the multilayer dressing has a considerable protective effect against HUs, as it improves the dissipation of tissue loads on a range of supports, by deformation of the dressing, which takes loads off from the tissues. We termed the reduction in volumetric exposures to effective strains as the ‘protective ratio’ of the multilayer dressing. Interestingly, this protective ratio was similar for the different support surfaces (e.g. ~3 for strains >0.5; Fig. 7b), which indicates that the biomechanical efficacy of the multilayer dressing is maintained on different (appropriate) supports and is particularly prominent in the more dangerous, high-end domain of strains. We attribute this quality to: (1) the mechanical cushioning provided by the soft layers of the multilayer dressing (layers II and IV); (2) the shear loads taken by the dressing due to the material transition effects between the soft (II, IV) layers of the dressing and the intermediate stiffer layer III; (3) the relatively low COF at the dressing-support interface. All these design features function together to minimize compression and shear deformations that develop in the skin and subdermal fat. The single-layer dressing was able to mildly reduce soft tissue exposure to elevated strains since it does provide some cushioning, however, lacking the above complex physical structure of the multilayer dressing, it could not, and did not provide the same extent of protection.

Since HUs often start internally and progress outwards, examining mechanical loads at the insertion of the AT should be an integral part of modeling studies in the context of HUs. Indeed, we found stress concentrations in the lateral aspect of the tendon insertion site, which were reduced due to the protective effect of the multilayer dressing, particularly under combined compression and shear.

The present work clearly shows that prophylactic use of multilayer heel dressings offers a biomechanical protective effect against HUs. Nevertheless, assumptions that are always involved in modeling inevitably introduce some limitations, which should be discussed. First, the anatomy can be different in the elderly and frail, where atrophy and loss of bone and soft tissues are
typical. Likewise, pathophysiological changes in the mechanical stiffnesses of soft tissues e.g. due to scars, edema and diabetes should affect the state of tissue loads. Additionally, as in most models of human tissues, the mechanical properties of tissues are mostly adopted from animal tissue data. Furthermore, refinements of the mechanical properties of the dressing materials can be introduced, as the current Poisson’s ratio is an effective value (for all the layers, taken together), though there are challenges in measuring Poisson’s ratios of thin structures.

In closure, we found that the use of a multilayer dressing on the heel efficiently reduces exposures of weight-bearing soft tissues to elevated strains and stresses, on a variety of supports, which should lower the risk for HUs. Indeed, independent RCTs recently confirmed the efficacy of the Mepilex® Border Heel design [19], and so, taken together, the modeling and RCTs indicate a great promise in taking this route for prevention.

**Conflict of interests**

None.

**Acknowledgments**

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**References**


The biomechanical efficacy of dressings


