A numerical study to analyse the risk for pressure ulcer development on a spine board

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A B S T R A C T

Background: Spine boards are used to immobilise accident victims suspected of having spinal injury. Guidelines about the maximum time patients remain on the board are often exceeded and on occasions may lead to pressure ulcers. Etiological research has shown that two processes ultimately lead to pressure ulcers: "ischemic damage" which takes several hours to initiate and "deformation damage" at high strains. The latter process is very quick and the first signs of cell damage are already evident within minutes. Thus in order to minimise the risk of pressure ulcer development during prolonged loading, a new soft-layered long spine board has been designed.

Methods: A subject specific numerical approach has been adopted to evaluate the prototype spine board in comparison to a conventional spine board, with reference to the estimated strains in the soft tissues adjacent to the sacrum in the supine position. The model geometry is derived from magnetic resonance images of three human volunteers in an unloaded situation. The loaded images are used to "tune" the material parameters of skin, fat and muscle. The prediction of the deformed contours on the soft-layered board is used to validate the model.

Findings: Comparison of the internal strains in muscle tissue near the spine showed that internal strains on the soft-layered board are reduced and maximum strains are considerably less than the threshold at which deformation damage is possible. By contrast, on the rigid spine board this threshold is exceeded in all cases.

Interpretation: The prototype comfort board is able to reduce the risk for deformation damage and thus reduces the risk of developing pressure ulcers.

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1. Introduction

The spine board is a device used for spinal immobilisation of accident victims suspected of having a spinal injury (Fig. 1a). This method of immobilisation is well established in pre-hospital and in-hospital trauma protocols (American College of Surgeons Committee on Trauma, 2004; National Association of Emergency Medical Technicians, 2006). Although existing guidelines advise the time spent on the spine board to be kept to a minimum (Brownlee, 2005; Vickery, 2001), in practice, patients remain on the board for prolonged periods, due to the on-scene treatment and evaluation in the emergency room and/or the radiological facility (Cooke, 1998; Stagg and Lovell, 2008). Since the spine needs to be immobilised during this period, the removal of patients from the spine board represents a low priority issue to the emergency room staff. Indeed the average time patients are reported to spend on a spine board has been estimated at 77 min (Lerner and Moscati, 2000; Sheerin and de Frein, 2007) although, in some cases it can exceed 2 h (Malik and Lovell, 2003; Yeung et al., 2006). Although stabilisation of the spine remains a critical requirement for trauma patients, it is also clear that prolonged immobilisation on the spine board causes pain and discomfort (Cordell et al., 1995; Hauswald et al., 2000; Zlupko et al., 2004) and, on occasions, may lead to the development of pressure ulcers (Baldwin and Ziegler, 1998; Cordell et al., 1995; Watts et al., 1998). All devices for pre-hospital care can be divided into two groups: those with a rigid surface and those with a softer padded surface. All currently available spine boards have a rigid surface. Some manufacturers have developed optional padding to soften the surface of the rigid spine board. The vacuum mattress, when deflated, also has a rigid surface. The soft-layered spine board is the only device with a built-in soft surface.

"A pressure ulcer (PU) is a localised injury to the skin and/or underlying tissue, usually over a bony prominence, as a result of pressure or pressure in combination with shear" (EPUAP/NPUAP, 2009). In addition to individual suffering, the development of PU’s leads to a prolonged hospital stay (Allman et al., 1999; Graves et al., 2005) and increased treatment costs (Dealey and Posnett, 2012). The incidence rate of PU’s in European hospitals is estimated to be between 18 and 20% (Nijs et al., 2008), and in the range of 10 to 18% in USA hospitals (Whittington...
and Briones, 2004). In the Netherlands, the incidence rate is estimated to be between 5 and 8% (when category 1 PU is excluded) in hospitals, and this value would inevitably be higher in home care and community settings (Hallens et al., 2012).

In the last decade considerable progress has been made in understanding the aetiology of PU’s. There is increasing consensus that, although the majority of pressure ulcers is initiated at the skin surface and may extend to deeper layers, there is also a category that starts in deeper soft tissue layers adjacent to bony prominences. In the 2009 guidelines for PU prevention and treatment (EPUIAP/NPUAP, 2009) the latter are referred to as Deep Tissue Injury (DTI). DTI is especially an issue for subjects that have impaired sensation, most notably spinal cord injured individuals, or unconscious patients. In this case the subject does not feel the mechanical-induced subcutaneous damage and it is not uncommon that once these wounds are identified they are already very severe.

It is hypothesised that superficial wounds mostly result from a combination of shear forces and pressures at the interface between skin and supporting surfaces, while DTI is a result of high stresses and strains adjacent to bone. It is proposed that spine boards could lead to DTI. Vulnerable areas for DTI with respect to the spine board are the sacrum, the heels, the shoulders and the back of the head.

In the last 15 years a variety of model systems have been used to study the aetiology of DTI, ranging from in vitro studies on single cells (Bouten et al., 2001; Wang et al., 2005), to tissue engineered muscle constructs (Breuls et al., 2003; Gawlita et al., 2007; Gefen et al., 2008) and animal studies (Bosboom et al., 2001; Ceelen et al., 2008; Loerakker et al., 2011; Stekelenburg et al., 2008). All these studies, focused on skeletal muscle cells and tissues, confirmed that two damage mechanisms play a major role in prolonged mechanical-induced damage. In most practical situations it is impossible to prevent the occlusion of blood vessels, resulting in partial or complete reduction of blood perfusion at mechanically loaded body sites. This leads to a state of ischaemia in the tissue, resulting in hypoxia. Accordingly, muscle tissue will change from an aerobic to anaerobic metabolism. This will cause an accumulation of metabolite waste products in the interstitial space, a reduction of the pH and eventually cell death. Healthy subjects will normally adopt pressure relieving strategies, involving changing positions, to minimise any risk of damage accumulation.

The second damage mechanism involves direct deformation damage at a threshold level, which for maximum shear strain has been estimated to be between 0.5 and 0.6 (50 to 60%) (Loerakker et al., 2011). The cause for this mechanism is not clear yet. It has been proposed that failure of the cell membrane occurs or there is direct disruption of the cytoskeletal elements. Nonetheless, all the model systems revealed a strong correlation between local tissue strain and damage development (Loerakker et al., 2011). It is also evident from studies with human volunteers that adjacent to bony prominences internal tissue strains often exceed the strain threshold level, during prolonged sitting (Linder-Ganz et al., 2007).

In the current paper the values of mechanical strains will be estimated in sacral tissues of subjects supported on a spine board. Two different support surfaces will be considered, namely, a standard long spine board and a prototype spine board with a soft-covered inlay, the soft-layered long spine board (Fig. 1b). A recent comparative study reported increased comfort and significantly lower interface pressures when measured on the soft-layered long spine board (Hemmes et al., 2010).

Two specific questions will be addressed:
1. Has the deformation threshold exceeded in the sacral area of subjects lying in supine position on a spine board?
2. Is the prototype soft-layered long spine board capable of reducing the internal strains to values below the deformation damage threshold?

To answer these questions, a combination of Magnetic Resonance Imaging and Finite Element modelling has been employed.

2. Methods

Magnetic Resonance Images of the sacral area of able bodied volunteers were recorded on both support surfaces. From these images a subject specific finite element model was developed. The model was tuned and validated on the undeformed and deformed images on the standard spine board and used to calculate internal tissue strains in the soft tissues near the sacrum.

2.1. Image acquisition

The geometry of the model was based on MR images of thirteen volunteers, with varying morphology, each scanned in a supine lying position on both the rigid spine board and the soft-layered long spine board. Images were also recorded with the subject lying in prone position in a standard MRI to depict the unloaded shape of the sacral region of each subject. Approval was obtained from the Institutional Review Board of the Maastricht University Medical Centre. Written informed consent was obtained from all volunteers before they were enrolled on this study.

After extensive inspection of all data sets, three representative subjects were selected to generate three anatomically accurate FE models.
meshes. Selection was based on a variation in anatomy and body mass index, visual accuracy of the data sets and whether subjects were lying aligned to the MRI scanner. The characteristics of the three female subjects are summarised in Table 1. The images in the prone position serve as an unloaded reference state of the buttocks area, on which the FE mesh is based. The images on the rigid spine board and the soft-layered long spine board were used to compare the output of the model. Specific focus of the model involved the most outward pointing prominences of the sacrum, the medial sacral crest, a few centimetres above the coccyx, and the femoral heads, as indicated in Fig. 2.

2.2. Mesh development

A subject specific two dimensional, plane stress, finite element model was developed of the selected subjects, based on the high resolution images close to the sacral area. The images that were obtained in the prone position, were used for the segmentation and considered to represent the undeformed state of the tissue area (Fig. 3a). Images were segmented in the software code Mimics (Materialise NV, Leuven, Belgium). After segmentation a mesh was developed with quadratic triangles and transferred to Marc Mentat 2008r1 (MSC Software).

2.3. Material properties

The model had four components to represent: bone, muscle, fat and skin tissues. The region in the mesh ascribed to bone was in fact tissue consisting of very stiff ligaments (Table 2). We assumed this would not affect the strains in the area of interest around the sacrum.

The skin, fat and muscle tissues were modelled with an incompressible, nonlinear Ogden material law with strain energy density $W$ (Ogden and Sgura, 2004):

$$W = \mu \left( \lambda_1^\alpha + \lambda_2^\alpha + \lambda_3^\alpha - 3 \right)$$

with $\lambda_{1,2,3}$ the principal stretch ratios and $\mu$ [kPa] and $\alpha [-]$ the material properties.

A series of parameter studies with the reference model (model II in Table 1) were performed to test the validity of the FE model. In particular the measurable contours of the deformed image on the rigid spine board were compared to the finite element predictions. First the mesh density was varied to identify the optimal compromise between computer time and accuracy. Close examination of the 2-D-model as a representation of a 3-D-subject revealed that a plane stress analysis, as opposed to a plane strain analysis, yielded the best fit on the MRI images. This suggests that tissues can deform easily in the direction perpendicular to the plane of the view. The friction coefficient between skin and supporting surface was varied and although this only marginally influences the contours, it did have a large influence on the calculated shear strains in the skin. Accordingly, a worst case scenario was chosen with a friction coefficient of 1.

The fitting was mostly influenced by the choice of the material parameter assigned to skin, fat and muscle. The image revealed a clear distinction between the tissue layers and thus manual fitting of each tissue was possible, leading to the material properties detailed in Table 1. The bone and stiff ligaments were modelled as an isotropic, linear elastic material with a high modulus of elasticity (3.4 [GPa]). This resulted in the mesh and material groups as shown in Fig. 4.

2.4. Boundary conditions

A contact option between skin and the supporting surface was used with Coulomb friction and a friction coefficient of 1. First, the support was moved upwards until contact was made between the skin and the surface. During this phase a central node in the model was fixed to prevent rigid body motion of the body. Secondly, a body force was applied to the body to simulate gravity. This body force was based on the density of water ($10^{-5}$ N/mm³). By integrating over the total area of the mesh, and taking the product of slice thickness (1 mm) the body force was assumed to approach the

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**Fig. 2.** a) Side view of the hip area. The red line a few centimetres above the coccyx is the region of interest. b) Front view of the hip area. The red line a few centimetres above the femur heads is the region of interest.

**Fig. 3.** a) High resolution MRI image made close to the sacral area in prone position (the image is rotated 180° for easy comparison with the supine image. b) Resolution MRI image made close to the sacral area in supine position.
were elements. Accordingly, the stiffness of the entire structure was represented by line elements acting as nonlinear, extensional springs that were fixed to a "virtual" line a few centimetres above the cushion. Again parameters were tuned in such a way that the outer contour in the loaded condition in the reference model matched the measured contours. In model I and II identical properties were used for the load sharing of the modelled slice at the sacral region of the adult human model.

The Spine board was modelled as a rigid surface. For the soft-layered long spine board a finite element model was built that include the cushion and the cover layer, which makes up its inlay (Fig. 5). Indentation tests were performed to measure the material properties of the cushion. The very large deformations, observed in the experiments on the spine board (up to 90%) led to convergence problems when using solid elements. Accordingly, the stiffness of the entire structure was represented by line elements acting as nonlinear, extensional springs that were fixed to a "virtual" line a few centimetres above the cushion. Again parameters were tuned in such a way that the outer contour in the loaded condition in the reference model matched the measured contours. In model I and II identical properties were used for the simulations.

### 3. Results

Fig. 6 shows the deformed images and the numerically determined contours of the model projected on the images of the 3 subjects on the rigid spine board. The model was "tuned" on the reference subject, with a focus on the area of interest, as indicated by the yellow dotted box in Fig. 6b. The same material parameter set was used for all three subjects, so the only difference for model I (Fig. 6a) and model III (Fig. 6c) is the subject specific geometry of the bones and tissues. The matching of the model to the images of the three subjects is reasonably accurate, suggesting little differences in the material properties of their individual tissues.

Fig. 7 shows the corresponding images and contours for the same subjects, lying on the soft-layered long spine board. It is interesting to note that for this case the reference model now provides the least accuracy, with a clear variance between the contours surrounding the fat layer and the MR images. By contrast, the contours derived from model I and III present a very good match with the images.

The estimated distribution of the maximum shear strains with the reference model on both the spine board and on the soft-layered long spine board reveals high strains, in excess of 50%, in the muscle and the fat around the sacrum on the rigid spine board. Such high strains are absent in the corresponding layers from the soft-layered long spine board (Fig. 8). Also high strains are found in the area around the ilium bones. These finding are not, however, considered to represent a realistic situation as the tissues around these bones are modelled with the same properties as muscle and the tissues cannot slide over the bone.

To enable a more quantitative comparison between the three subjects and between the spine board and the soft-layered long spine board a region of interest (ROI) around the sacrum is selected, depicted by region A in Fig. 4. Within the rectangular ROI, the relative area corresponding to different strain ranges was plotted for all cases. Fig. 9a depicts the histogram for the area around the sacrum for maximum shear strains ranging from 0 to 0.8. The shaded area in the range between 0.5 and 0.6 corresponds to the threshold range where damage due to deformation has been shown to be initiated (Loerakker et al., 2010, 2011). Fig. 9b magnifies the estimated strains between 0.3 and 0.8. It is very clear that for all subjects the spine board yields strains within the critical strain range, while on the soft-layered long spine board all maximum shear strains are less than 0.5.

### 4. Discussion

Aetiologial research has established that in skeletal muscle, in addition to cell death as a result of ischaemia, a different damaging process is evident, which has been termed "direct deformation damage". This process is associated with a very strong correlation between the local levels of strain applied to tissues and the amount of damaged observed in both animal studies and in vitro studies. Although the exact mechanisms are not fully clear, it is known that below a maximum shear strain threshold of between 0.5 and 0.6 (50 to 60%) this damage will not occur, whereas when this threshold is exceeded the first signs of cell damage are evident for periods as short as 10 min (Loerakker et al., 2011). The time it takes from these early signs to develop a full wound is not known, but it seems prudent to avoid this threshold strain value. Recent research has shown that ischaemic-induced damage can occur (Berg et al., 2010), although its effect is registered only after several hours (Loerakker et al., 2011).

In all three subjects strains of up to 0.8 were found in muscle as well as fat around the sacral bone, suggesting the use of the rigid spine board can result in direct deformation damage.

It is also clear from the FE estimations that for each of the three subjects the maximum strains on the soft-layered long spine board did not exceed the deformation damage threshold. This implies that transportation of patients on a soft-layered long spine board is fairly safe, at least in terms of preventing deformation damage. However, it should be noted that for such patients presenting factors such as trauma, shock and unconsciousness, the threshold strains to avoid damage will inevitably be lower. Although ischaemia is a slow process, the times that patients are maintained on the spine board are in some cases prolonged and thus consideration of individual patient sensitivity is critical. Furthermore, the present analysis was only focussed on the sacral area and other studies in our group, involving

**Table 2**

<table>
<thead>
<tr>
<th>Tissue</th>
<th>E [kPa]</th>
<th>α</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skin</td>
<td>8</td>
<td>5</td>
</tr>
<tr>
<td>Fat</td>
<td>1</td>
<td>5</td>
</tr>
<tr>
<td>Muscle</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Bone</td>
<td>3.4 * 106</td>
<td>0.36</td>
</tr>
</tbody>
</table>

**Fig. 4.** The finite element mesh for the reference model, with the different material groups used.
blood gas tension levels and pressure measurements, have indicated that the shoulder area is particularly vulnerable on the spine board (unpublished data).

We only examined 3 participants and these people were healthy. The stresses exerted onto the tissues due to tissue deformation are likely to be independent of underlying diseases. However, the effect of this deformation on the tissues may very well be influenced by the pre-existent conditions of the patient (e.g. diabetes). In this study we only investigated the effect of different surface properties on the extent of deformation. The level of damage by this deformation was beyond the scope of our study.

In addition, the model is associated with a number of limitations. For example it is a two-dimensional plane stress model and does not take into account tissue deformation in the direction perpendicular to the modelled cross-section. Although, in principle, it is possible to create 3-D whole body models, such a process is elaborate and time-consuming process and it is difficult to produce patient specific models. In a similar manner the material parameter values used in the model for
skin, fat and muscle tissues are only based on average values of limited sample size. On the other hand, all models were analysed with only one data set based on tuning the reference model on the rigid spine board. The good agreement between predicted and measured contours with one single data set provides some confirmation of validity of the employed methodology.

**Conclusions**

In the sacral area of all three subjects with BMI’s ranging from 19 to 27 maximum shear strains higher than 0.55 were found. These strains may lead to direct internal deformation damage in muscle within minutes. For each of the three subjects the maximum strains on the soft-layered long spine board did not exceed the deformation damage threshold.

**Conflict of interest statement**

The authors have no conflict of interest concerning the contents of this paper.

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