Pressure Pathogenesis of DTI

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Identifying risk factors – a step towards understanding the pathogenesis of DTI

and there are also practical reasons to know risk factors

- Identify an individual’s level of risk at admission
- Identify priorities of care
- Establish baseline for future reference (especially if complications arise)
- Make an appropriate use of (ever-limited …) resources
Immobility is the most important risk factor

Protocol for determining postural change patterns of healthy individuals who sit in a wheelchair:

- Each subject was sitting in a wheelchair for 90 minutes, during which he watched a movie on computer #2
- During that time, interfacial pressures and motion data were recorded on computer #1

Linder-Ganz et al., Technology and Healthcare 2007
Stick diagrams of frontal plane movement

Time = 323 Sec
Sh = 5°
Th = 12°
Lu = 12°

Time = 1160 Sec
Sh = 5°
Th = 5°
Lu = 20°

Time = 1834 Sec
Sh = 6°
Th = 7°
Lu = 8°

Time = 2182 Sec
Sh = 5°
Th = 11°
Lu = 14°

Time = 3790 Sec
Sh = 11°
Th = 8°
Lu = 14°

Time = 4194 Sec
Sh = 7°
Th = 12°
Lu = 9°

Before movement
During movement
End of movement
Sh = shoulders segment angle
Th = thoracic-spine segment angle
Lu = lumbar-spine segment angle

Linder-Ganz et al., Technology and Healthcare 2007
Healthy individuals move frequently!

Table 1. Time characteristics of motion during wheelchair sitting. Data were acquired from 10 normals (5 males and 5 females, age = 28±3 years, weight = 65±9 Kg, and height = 174±9 cm) during a timeframe of 70 minutes within a 90 minutes experiment.

<table>
<thead>
<tr>
<th># of Subjects</th>
<th>Gender</th>
<th>Number of pressure relief movements</th>
<th>Intervals between movements [Minutes]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Frontal</td>
<td>Sagittal</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Left</td>
<td>Right</td>
</tr>
<tr>
<td>5</td>
<td>Female</td>
<td>4.8±4.6</td>
<td>9±6.1</td>
</tr>
<tr>
<td>5</td>
<td>Male</td>
<td>0.4±0.9(^a)</td>
<td>8±4.7(^a)</td>
</tr>
<tr>
<td>Pooled for Gender:</td>
<td>2.6±3.9(^a)</td>
<td>8.5±5.2(^a)</td>
<td>11.1±5.9</td>
</tr>
</tbody>
</table>

\(^a\) Number of movements to the right in the frontal plane was significantly greater than those to the left (p < 0.05).

On average, subjects moved every 6 minutes in the frontal plane, and independently, every 6 minutes in the sagittal plane.
But, extents of movements are mild ...

Maximal change of angle during movement:

- 10 subjects (5 M, 5 F)
- 90 min trials
- 70 min of data acquisition

8 ± 4
14 ± 7
15 ± 7

Body segment

Linder-Ganz et al., Technology and Healthcare 2007
Deformations across the different hierarchical scales in weight-bearing soft tissues

Shoham and Gefen, *J Tissue Viability* 2012
Shabshin et al., *JRRD* 2010
Sustained tissue deformations cause cell distortions, and in particular - stretching of the plasma membrane.

The plasma membrane becomes more permeable when substantially stretched.

Ben-Or et al., *Journal of Biomaterials and Tissue Engineering* 2013
Which factors can affect local tissue and cell deformation levels – and therefore the risk for DTI?
Internal tissue loads are support-dependent
Development of tissue loads over time is also support-stiffness dependent.
Skin and subcutaneous tissue loads as function of the cushion stiffness

[Graphs showing accumulated strain dose and stress rate at initial contact for different cushion stiffness values (5, 25, 100 kPa) for young and aged.]
Studying the effect of the body mass on deep tissue loads

Models representing anatomical and loading changes in the buttocks, associated with bodyweight:

- % body fat = 8.9%
  BMI = 15.4 kg/m²

- % body fat = 15.1%
  BMI = 17.7 kg/m²

- % body fat = 18.4%
  BMI = 19 kg/m²

- % body fat = 30.3%
  BMI = 24.5 kg/m²

- % body fat = 38.3%
  BMI = 29.3 kg/m²

- % body fat = 47.8%
  BMI = 40 kg/m²

Elsner & Gefen, *Journal of Biomechanics* 2008
Sopher et al., *Journal of Biomechanics* 2010
There appears to be an optimal range of BMI within which intramuscular loading during sitting is minimal.
Studying the effect of intramuscular fat infiltration

Mild fat infiltration

Moderate fat infiltration

Severe fat infiltration

Sopher et al., *Journal of Biomechanical Engineering* 2011
Studying the effect of intramuscular fat infiltration (cont.)

The intensity of loading in muscle tissue increases with the increase in severity of fat infiltration.

Sopher et al., *Journal of Biomechanical Engineering* 2011
Studying the effect of skin scars
Studying the effect of deep scars or deep scars in combination with superficial scars

Hourglass-shaped scarring

Sandwich-shaped scarring

Deep muscle scarring

Superficial scarring

Sopher et al., *Journal of Biomechanical Engineering* 2011
Studying the effect of scars (cont.)
Studying the effect of scars (cont.)

Muscle tissue loads increase with an increase in the size of scars (especially for hourglass-shaped and sandwich-shaped types).

Sopher et al., Journal of Biomechanical Engineering 2011
DTI in the heels
Fig. 1 – The finite element (FE) model of the heel: (a) A sketch of a foot resting on a horizontal support with the heel region of interest modelled herein marked with a cube. The red arrow further marks the plane of view used in the other parts of this figure. (b) Three-dimensional geometry of tissue components in the FE model. (c) The model configurations simulating a foot for which the lateral aspect is resting at 60° (foot abduction; left panel) and 90° (foot upright; right panel) with respect to the support, and the corresponding FE meshes.
An inclined foot posture puts a bedridden patient at a higher risk for heel ulcer with respect to an upright foot posture.
DTI in the heels (cont.)

Fig. 3 – Peak values of (a) principal compressive strain, (b) principal tensile strain, (c) maximal shear strain, and (d) strain energy density in the heel fat pad, plotted versus the elastic modulus of the support, for the 60°-oriented and 90°-oriented heel simulation datasets.

Fig. 5 – Percentage of heel fat pad volume which exceeds critical load levels of (a) strain energy density = 0.3, 0.5 or 0.7 kPa, and (b) von Mises stress = 1.5, 2 or 2.5 kPa, plotted versus the elastic modulus of the support, for the 60°-oriented and 90°-oriented heel simulation datasets.
MRI studies of the posterior heel in the supported foot support the computational findings.

<table>
<thead>
<tr>
<th>90°-oriented foot</th>
<th>60°-oriented foot</th>
</tr>
</thead>
<tbody>
<tr>
<td>5.86 mm</td>
<td>3.91 mm</td>
</tr>
<tr>
<td>10.55 mm</td>
<td>9.77 mm</td>
</tr>
</tbody>
</table>

44% strain

60% strain
Heel support devices (i.e. extrinsic factors) can also be evaluated.

<table>
<thead>
<tr>
<th>Heel placed on a stiff surface</th>
<th>Heel support #1</th>
<th>Heel support #2</th>
<th>Heel support #3</th>
<th>Non-weight bearing</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.91 mm</td>
<td>5.08 mm</td>
<td>4.69 mm</td>
<td>7.03 mm</td>
<td>9.77 mm</td>
</tr>
</tbody>
</table>

Strain:
- Heel placed on a stiff surface: 60%
- Heel support #1: 48 - 52%
- Heel support #2: 28%
Understanding the pathogenesis and risk factors allows development of technologies for prevention
Gaps in current science/technology related to DTI

- A need for patient-specific analyses of internal tissue loads that can be carried out continuously in time, desirably in real-time

- Analyses should be sensitive to the patient’s anatomy and momentary posture, as well as to the wheelchair/cushioning used

Such technology would then allow alerting on over-exposure to internal tissue loads in a clinical setting, at a nursing home or at home, by miniaturization into a portable monitor device.
How our system works
Results: Group comparisons of internal gluteal muscle loads (i)

- **Compression stress**:
  - Controls: 6.8 ± 5 kPa
  - Subjects with SCI: 26.3 ± 11.8 kPa
  - ~4-times higher in the SCI group

- **Tension stress**:
  - Controls: 3.2 ± 2.5 kPa
  - Subjects with SCI: 9.8 ± 5.8 kPa
  - ~4-times higher in the SCI group

- **von Mises stress**:
  - Controls: 6.3 ± 4.5 kPa
  - Subjects with SCI: 24.5 ± 10.1 kPa
  - ~4-times higher in the SCI group

- **Shear stress**:
  - Controls: 3.2 ± 2.5 kPa
  - Subjects with SCI: 10.8 ± 4.2 kPa
  - ~5-times higher in the SCI group

* *p < 0.05
Subjects with SCI

3.5 ± 2.2

Number of internal tissue load relieves performed during the 90 min trial

~35-times higher in the SCI group

1,190 ± 407

Controls

41,635 ± 31,152

Subjects with SCI

~50-times higher in the SCI group

13,692 ± 10,240

Controls

35.5 ± 5.6

Subjects with SCI

* Internal tissue load relief = peak gluteal m. compression stress reading below 2 kPa for at least 1s

*p < 0.05

Linder-Ganz et al., Annals of Biomedical Engineering 2009
Discussion: Findings from the real-time system

• Significantly higher stresses and stress doses were found in gluteal muscles of patients with SCI compared with controls.

• The present real-time method/system can be applied in the future to several potential medical applications, e.g.:
  
  (i) Alarm system to prevent DTI
  (ii) Wheelchair/cushion adjustments
  (iii) Design of wheelchairs/cushions
  (iv) Feedback during rehabilitation program

Though an important limitation is computational resources for running the FE analyses in real-time.
Analytical approaches allow cost-effectiveness in prevention

Assuming no slip at bone-muscle interface:

\[
\sigma_y = -\frac{c_y}{R h} \left( r^2 - \frac{r^2}{R^2} \right) \quad \text{for } r \in [0, a]
\]

\[
\sigma_\theta = -\frac{c_\theta}{R h} \left( r^2 - \frac{r^2}{R^2} \right) \quad \text{for } r \in [0, a]
\]

\[
\sigma_z = -\frac{c_z}{R h} \left( r^2 - \frac{r^2}{R^2} \right) \quad \text{for } r \in [0, a]
\]

where

\[
\sigma = \left( \frac{4K_3}{\pi} \right)^{1/3}
\]

Assuming free slip at bone-muscle interface:

\[
\sigma_y = 0
\]

\[
\sigma_\theta = 0
\]

\[
\sigma_z = 0
\]

Mechanical properties of tissues:

\[
\epsilon_{11} = \epsilon_{33} = \frac{(1-\nu)\alpha E}{(1+\nu)(1-2\nu)}
\]

\[
\epsilon_{33} = \epsilon_{11} = \frac{\nu\alpha E}{(1+\nu)(1-2\nu)}
\]

\[
\epsilon_{44} = \frac{\alpha E}{2(1+\nu)}
\]

Solution for the bone-muscle contact stress:

\[
\sigma_{33}(t, r, z) = \beta \left( \sigma_{33}^{\text{base}}(t, r, z) \right) + (1-\beta) \left( \sigma_{33}^{\text{max}}(t, r, z) \right)
\]

Fig. 5 An example of the a loads, b peak internal von Mises stress, and c internal stress dose under the left and right ischial tuberosities of subject D3 (Table S2)
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