The Compression Intensity Index: A practical anatomical estimate of the biomechanical risk for a deep tissue injury

Amit Gefen
Department of Biomedical Engineering, Faculty of Engineering, Tel Aviv University, Tel Aviv 69978, Israel
Tel.: +972 3 640 8093; Fax: +972 3 640 5845; E-mail: gefen@eng.tau.ac.il

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Abstract. Pressure-related deep tissue injury (DTI) is a severe form of pressure ulcer that initiates in compressed muscle tissues under bony prominences, and progresses superficially towards the skin. Patients with impaired motosensory capacities are at high risk of developing DTI. There is a critical medical need for developing risk assessment tools for DTI. A new anatomical index, the Compression Intensity Index: $CII = (BW/Rt)^{1/2}$, which depends on the body weight ($BW$), radius of curvature of the ischial tuberosities ($R$) and thickness of the underlying gluteus muscles ($t$), is suggested for approximating the loading intensity in muscle tissue during sitting in permanent wheelchair users, as part of a clinically-oriented risk assessment for DTI. Preliminary CII data were calculated for 6 healthy and 4 paraplegic subjects following MRI scans, and data were compared between the groups and with respect to a gold standard, being a previously developed subject-specific MRI-finite-element (MRI-FE) method of calculating muscle tissue stresses (Linder-Ganz et al., J. Biomech. 2007). Marked differences between the $R$ and $t$ parameters of the two groups caused the CII values of the paraplegics to be $\sim 1.6$-fold higher than for the healthy ($p < 0.001$), thereby indicating on the sensitivity of this parameter to the pathoanatomical changes that occur in the buttocks with paraplegia. Data of CII correlated reasonably with the gold standard calculations of MRI-FE muscle stresses (correlation coefficient 0.65). Since CII measurements do not require highly-specialized biomechanical numerical analyses such as MRI-FE, CII has the potential to serve as a practical, quick, and cost-effective approximation of the loading intensity in muscles of wheelchair-bound or bedridden patients. Hence, CII measurements can be integrated into DTI-risk-assessment tools, the need of which is now being discussed intensively in the American and European Pressure Ulcer Advisory Panel meetings.

Keywords: Pressure ulcer, decubitus, bed sore, analytical model, contact mechanics

1. Introduction

Deep tissue injury (DTI) is a recently defined medical condition of a pressure-related injury to subcutaneous tissues under intact skin, due to prolonged compression of bony prominences on underlying soft tissues, particularly muscles [1,2,13]. It is one of the severe forms of pressure ulcers, and is characteristic to patients with impaired motosensory capacities such as paraplegics and quadriplegics, in whom it may cause serious complications e.g. a myocardial infarction, renal failure, sepsis and eventually death [2]. Very often, DTI develops under the ischial tuberosities (IT) where underlying gluteus muscle tissues of chronically sitting or lying patients sustain prolonged, excessive compression by the IT (Fig. 1a).
DTI is a particularly risky condition because it may develop unnoticeably by the patient and medical caretakers, or show inconclusive symptoms such as discoloration of the skin that can also be caused by superficial pressure ulcers or incontinence lesions [2]. This inherent problem in diagnosis of DTI is evident in its recent definition of a “suspected DTI” by the US National Pressure Ulcer Advisory Panel (NPUAP, http://www.npuap.org/): “purple or maroon localized area of discolored intact skin or blood-filled blister due to damage of underlying soft tissue from pressure and/or shear. The area may be preceded by tissue that is painful, firm, mushy, boggy, warmer or cooler as compared to adjacent tissue.” Accordingly, the problem of DTI attracts growing attention of both the clinical and basic research communities, and the problems of risk assessment of DTI are discussed intensively in the meetings of the NPUAP and its European equivalent, the EPUAP, over the last years [4]. Along with the increasing understanding of the etiology of DTI from a basic research aspect [5,10,11,13,14,23,24], there is a critical need to develop medical, practical risk assessment tools, for evaluating whether an individual should be closely monitored, or receive prevention means, for minimizing the risk for DTI. While there are general tools available for assessment of a pressure ulcer risk, e.g. the Braden Scale, these may not necessarily predict the risk for a DTI, because DTI was not considered in the development of all currently available risk assessment tools. Also, while there are some sophisticated, finite element (FE) based methods for accurately calculating the mechanical conditions in muscle tissues under the IT during supported postures [23,24], they require high resources and considerable expertise that may limit a wide clinical use.

The purpose of this communication is therefore, to suggest a practical, clinically-oriented and easy-to-measure anatomical index for a quick assessment of the mechanical loading intensity in muscles under the IT of an individual, and therefore, of the relative biomechanical risk for that individual to develop DTI, based on the well-established association between the magnitude of mechanical loads and extent of tissue damage [10,11,22,29]. This anatomical measure, the Compression Intensity Index (CII), is calculated from the ratio of body weight ($BW$) over the product of the IT radius of curvature ($R$) and gluteus muscle thickness ($t$), that can both be measured with a variety of imaging modalities, including MRI examinations (usually prescribed to paraplegic and quadriplegic patients as a medical routine) and ultrasound [20,21,24,25,27,30]. A theoretical proof-of-concept that the CII parameter is a measure of the muscle loading state is first provided, based on the general theory of contact mechanics. Second, preliminary CII data collected from small groups of healthy subjects and paraplegic patients using MRI are analyzed and compared, to demonstrate feasibility that CII is sensitive to the pathoanatomical changes in the IT and gluteus structures in paraplegia, and that it can reasonably approximate the intensity of muscle loading in individuals from both groups.

2. Methods

2.1. Theory

The pathoanatomy of the buttocks in paraplegic and quadriplegic patients has been documented in numerous studies (see [17] for a review). It includes gradual loss of muscle thickness [6,7] as well as cortical bone [8,28], which affects the “sharpness” (radius of curvature) of the IT. The effects of these structural changes on the muscle loading state can be evaluated, as a first approximation, using the contact mechanics theory. For this purpose, the IT-muscle contact problem during sitting or lying is represented as indentation of a rigid, frictionless half-sphere (“IT”) into an isotropic elastic layer (“muscle”) that is free to slide on a rigid foundation (“support surface”) when indented (Fig. 1b). The detailed, complete
Fig. 1. The problem of contact between the ischial tuberosity (IT) and gluteus muscle during sitting or lying: (a) Schematic drawing of the relevant anatomy during sitting. (b) Representation of the IT-muscle contact problem using the contact mechanics theory, where the IT is considered as a half-sphere with radius \( R \) that indents an elastic flat muscle layer with thickness \( t \), to a depth \( \delta \). (c) Examples of the distribution of compression strains \( \varepsilon(r) \) and contact pressures \( p(r) \) in the “muscle” at a distance \( r \) from the axis of indentation, for a parameter set \((R, t, \delta)\) that represents the means for the healthy subject group (in the calculation of stress distribution, using Eq. (2), an elastic modulus of 25.5 kPa was assumed for the muscle, based on Palevski et al. [26].

Closed-form set of solutions for the strain and stress distributions under the half-sphere indenter, for cases of both anisotropic and isotropic elastic layers, were recently provided for this contact problem by Ning and colleagues [25], who extended a previous work of Jaffar [18]. Assuming that the axial compressive strain beneath the IT is uniformly distributed through the thickness of the elastic muscle layer, the distribution of compression strains \( \varepsilon_{zz}(r) \) at a radial distance \( r \) from the axis of rotational symmetry is [18]:

\[
\varepsilon_{zz}(r) = -\frac{1}{t} \left( \delta - \frac{r^2}{2R} \right)
\]

where \( t \) is the thickness of the muscle layer, \( \delta \) is the depth of indentation of the IT into the muscle layer and \( R \) is the radius of curvature of the IT (Fig. 1c). For the isotropic layer case, the contact pressure distribution under the IT, \( p(r) \), is (Fig. 1c):

\[
p(r) = \frac{E}{1-\nu^2} \left( \frac{a^2 - r^2}{2Rt} \right)
\]

where \( E \) is the elastic modulus of the muscle layer (non-contracted muscle tissue can be assumed to behave elastically at long terms relevant to DTI formation, as demonstrated by Palevski and colleagues [26]), \( \nu \)
is the muscle’s Poisson’s ratio, and \( a \) is the radius of contact between the IT and muscle layer, calculated as [25]:

\[
a = \left[ \frac{4Prt(1-\nu)^2}{\pi E} \right]^{1/4} \approx \sqrt{2R\delta}
\]

(3)

where \( P \) is the compressive force of indentation, that is, the force transferred through the IT. Hence, the force transferred through the IT relates to the depth of indentation of the IT into muscle tissue, as [25]:

\[
P \approx \frac{\pi Er\delta^2}{t(1-\nu)^2}
\]

(4)

The peak contact pressure is obtained underneath the IT, on the symmetry axis at \( r = 0 \):

\[
p_{\text{max}} = p(r = 0) = \frac{Ea^2}{2(1-\nu)^2Rt}
\]

(5)

Considering that muscle tissue contains \( \sim 75\% \) water, incompressibility \( (\nu = 0.5) \) is assumed, and Eq. (3) is then substituted into eq. (5), which yields:

\[
p_{\text{max}} = \frac{2E}{3Rt} \left( \frac{Prt}{\pi E} \right)^{1/2}
\]

(6)

It is also possible to formulate \( p_{\text{max}} \) in terms of the indentation of the IT into the muscle \( (\delta) \) by substituting Eq. (4) into Eq. (6) under the assumption of muscle incompressibility:

\[
p_{\text{max}} \approx \frac{4\delta}{3t}E
\]

(7)

However, measuring \( \delta \) in individuals requires sophisticated MRI or ultrasound techniques [21,23,24], and so, Eq. (6) is a more practical form to implement. Specifically, since the IT indentation force \( P \) is linearly proportional to the body weight \( (BW) \) [9], the proportionality between the maximal contact pressure under the IT \( (p_{\text{max}}) \) and the anatomical parameters of an individual can be formulated, based on Eq. (6), as:

\[
p_{\text{max}} \propto \sqrt{\frac{BW}{Rt}}
\]

(8)

With implication to the DTI problem, Eq. (8) indicates that maximal tissue loading under the IT \( (p_{\text{max}}) \) is expected to be higher in individuals who either have IT with smaller radii of curvature \( (R) \), or who have thinner muscle thickness \( (t) \). To put this result in the context of pathoanatomical changes that may occur in the buttocks with paraplegia or quadriplegia, Eq. (8) further predicts that a combination of an especially sharp IT \( (\text{i.e. small } R) \) and thin, atrophied muscle \( (\text{low } t) \) is expected to increase intramuscular loading under the IT, which is a biomechanical risk factor for DTI. Equation (8) also shows that peak muscle loading is expected to increase with \( \sqrt{BW} \). Obesity has been long known as a risk factor for pressure ulcers, and has recently also been recognized as a risk factor for DTI [3]. Unfortunately, patients with paraplegia and quadriplegia, e.g. due to spinal cord injury (SCI), are more likely to be overweight and obese [31]. Last, an equivalent analysis that assumes an elastic muscle layer that is bonded to the rigid foundation (detailed in [25]), that is, to describe a condition of high friction with the seat, yields the
same proportional relation provided in Eq. (8). Hence, based on Eq. (8), a personalized, anatomy-based Compression Intensity Index (CII) can be defined:

\[ CII = \left( \frac{BW}{Rt} \right)^{1/2} \]  

The CII is a first approximation of the relative mechanical loading intensity in the gluteus of individuals, and does not require complex numerical (FE) analyses.

2.2. Preliminary measurements of the Compression Intensity Index

In order to test whether CII is indeed sensitive to the anatomical changes observed in paraplegia, a feasibility study was performed. All subjects recruited for this study were chosen for their previous participation in subject-specific MRI-FE studies previously conducted in our laboratory [23,24]. The MRI scans for measuring \( R \) and \( t \) were approved by the Helsinki committee of Sheba Medical Center (Ramat-Gan, Israel), where patients were studied, and informed consent was obtained from each participant.

The sample group included 6 healthy and 4 paraplegic subjects. Exclusion criteria for paraplegics were: an active pressure ulcer, pregnancy, no muscle tissue detected in the MRI scan, professional athletic activity, and inability to lift the body off the wheelchair using the arms. The \( R \) and \( t \) parameters of each subject were provided by MRI coronal scans of the buttocks in a non-weight-bearing sitting posture, conducted while subjects sat erect on a rubber tire placed on a specially-designed Perspex chair within an Open-MRI, as described in detail in [23]. Briefly, we measured \( R \) and \( t \) in a coronal scan showing both ITs, in which the soft tissue thickness under the ITs was minimal. We averaged measurements of IT radius (\( R \)) and gluteus thickness (\( t \)) from the two body sides per each subject, and calculated the CII of each subject using the mean \( R \), \( t \) data. We then compared the CII data of the two groups, and plotted them against previous subject-specific peak compression stress data obtained for the same subjects using our MRI-FE method [23,24] to determine (i) whether CII can reflect the anatomical differences between the healthy and paraplegics and (ii) if CII indeed predicts the loading intensity in the gluteus, as provided by the “gold-standard” MRI-FE method of Linder-Ganz et al. [23]. Statistical 2-tail, unpaired t-tests were used for all group comparisons, assuming unequal variances, and a \( p \) value > 0.05 indicated significance.

3. Results

The measurements of IT radius of curvature (\( R \)) and gluteus thickness (\( t \)) during non-weight-bearing sitting and the corresponding CII data are provided in Table 1 for all the healthy (\( N = 6 \)) and paraplegic (\( N = 4 \)) subjects. The \( R \) for the paraplegic group (mean ± standard deviation: 19.4 ± 2.5 mm) was \( \sim \)1.6-fold higher than for the healthy (12 ± 2.3 mm) (\( p < 0.01 \)), indicating a significant increase in IT radius of curvature in the paraplegics, likely due to cortical bone loss and/or shape adaptation of the bones to the sustained sitting loads. The \( t \) for the paraplegic group (5.6 ± 1.3 mm), was however, less than third the thickness in healthy (20.7 ± 2.4 mm) (\( p < 0.001 \)), which demonstrates the substantial loss of muscle tissue in paraplegia. Body weights did not differ significantly between the two groups (range 55–80 kg), but the marked differences between the \( R \) and \( t \) parameters caused the CII index of the paraplegics to be \( \sim \) 1.6-fold higher, on average, than for the healthy (\( p < 0.001 \), Fig. 2a). Hence, we conclude, based on these preliminary data, that the CII is sensitive to the pathoanatomical changes in the buttocks that are characteristic to paraplegia.

Next, we tested whether CII, being an anatomical index, is able to reflect the mechanical loading state in the gluteus, at least as a first approximation. Hence, we plotted the CII values of all 10 subjects...
Table 1

Anatomical data of participants. ITs = Ischial tuberosities

<table>
<thead>
<tr>
<th>Subject code</th>
<th>Gender / Age [years]</th>
<th>Body weight (BW) [kg]</th>
<th>Spinal level of injury / Time post spinal injury [years]</th>
<th>Mean radius of curvature of the ITs (R) [mm]</th>
<th>Mean thickness of the gluteus muscles under the ITs during sitting (t) [mm]</th>
<th>Compression Intensity CI = (BW/Rt)(^{1/2})</th>
</tr>
</thead>
<tbody>
<tr>
<td>Healthy 1</td>
<td>M/28</td>
<td>85</td>
<td>N/A</td>
<td>11</td>
<td>23</td>
<td>0.580</td>
</tr>
<tr>
<td>Healthy 2</td>
<td>F/27</td>
<td>60</td>
<td>N/A</td>
<td>8.5</td>
<td>18</td>
<td>0.626</td>
</tr>
<tr>
<td>Healthy 3</td>
<td>M/27</td>
<td>78</td>
<td>N/A</td>
<td>12</td>
<td>19.5</td>
<td>0.577</td>
</tr>
<tr>
<td>Healthy 4</td>
<td>F/29</td>
<td>55</td>
<td>N/A</td>
<td>11.5</td>
<td>24</td>
<td>0.446</td>
</tr>
<tr>
<td>Healthy 5</td>
<td>M/27</td>
<td>90</td>
<td>N/A</td>
<td>14.5</td>
<td>20.5</td>
<td>0.550</td>
</tr>
<tr>
<td>Healthy 6</td>
<td>F/26</td>
<td>55</td>
<td>N/A</td>
<td>14.5</td>
<td>19</td>
<td>0.447</td>
</tr>
<tr>
<td>Paraplegic 1</td>
<td>F/36</td>
<td>70</td>
<td>T3/17</td>
<td>22</td>
<td>6.5</td>
<td>0.700</td>
</tr>
<tr>
<td>Paraplegic 2</td>
<td>F/26</td>
<td>65</td>
<td>T3-4/3</td>
<td>17</td>
<td>4.5</td>
<td>0.922</td>
</tr>
<tr>
<td>Paraplegic 3</td>
<td>M/21</td>
<td>90</td>
<td>T6/1</td>
<td>21</td>
<td>4.5</td>
<td>0.976</td>
</tr>
<tr>
<td>Paraplegic 4</td>
<td>M/21</td>
<td>80</td>
<td>T3-4/2</td>
<td>17.5</td>
<td>7</td>
<td>0.808</td>
</tr>
</tbody>
</table>

N/A = Not applicable.

against the peak principal compression stress data we obtained previously for the same subjects using the subject-specific MRI-FE method of analysis [23,24], which is considered herein as the “gold standard” method. The plot, in Fig. 2b, shows a clear distinction between the paraplegic and healthy groups, and also indicates on a reasonable correlation (correlation coefficient 0.65) between the CII index and the “gold standard” peak compression stress data.

4. Discussion

The present communication suggests the Compression Intensity Index as a practical, easy-to-measure parameter that reflects the relative loading intensity in muscle tissue under the IT during chronic sitting, without the need to conduct complex patient-specific numerical simulations (e.g. as in [23,24]). This is of great medical benefit for fast, cost-effective routine screening of patients that may be at risk for DTI.

The preliminary CII data indicates on feasibility of the technique, and particularly, that CII, although based purely on anatomical measures, fairly approximates the mechanical loading intensity in muscles as measured using sophisticated, labor intensive numerical methods (Fig. 2b). If measurements of \( R \) and \( t \) will be obtained by means of ultrasound [20,21,27,30] rather than with MRI, CII measurements can be integrated into a cost-effective risk assessment for DTI, in facilities such as nursing homes, community clinics, small hospitals and rehabilitation centers where even in the future, it may appear impractical or too expensive to conduct patient-specific MRI-FE analyses (as suggested by Linder-Ganz and colleagues [23,24]). Naturally, following this present “proof-of-concept” study, larger-scale clinical studies, employing ultrasound (the study of Lin and colleagues [21] already proved feasibility of ultrasound measurements of \( R \) and \( t \); see Fig. 3 in their paper), are now needed to collect more data on the behavior of the CII in healthy and patients. Specifically, there is a need to determine how CII is influenced by gender, age, time post-SCI, and physical activity. Follow-up patient studies are very important in this regard, to determine whether there are threshold values of CII that can predict the likelihood for DTI.

Interestingly, bone loss at the IT in the paraplegic group, which significantly increased their \( R \), appears to counteract the loss of muscle thickness \( t \) in this group, as manifested in the product \( Rt \) in the denominator of the CII (Eq. (9)), however, the reduced \( t \) is substantially more dominant in the paraplegics, so that their CII is overall significantly increased with respect to healthy. This agrees well with the recent
Fig. 2. The compression intensity index (CII): (a) Comparison between healthy and paraplegic CII data ($p < 0.001$), and (b) Correlation of CII with peak muscle compression stress data calculated using a subject-specific MRI/finite element analysis for the same individuals, in previous studies [23,24]. Data points of the healthy group (circles) are grouped by a dashed line, and data points of the paraplegic groups (squares) are grouped by a solid line, to depict that the values for the two groups are distinct on the plot.

Patient-specific MRI-FE analyses of Linder-Ganz et al. [24] who demonstrated significantly higher strains and stresses in the gluteus muscles of sitting paraplegics compared with sitting healthy individuals.

A debate exists in the literature regarding obesity and pressure ulcers: Biomechanical studies of seat interface pressures generally report the highest pressures in thin individuals [19]. However, epidemiological studies indicate that among individuals with SCI, obesity is common [31], and also, that obesity is a risk factor for pressure ulcers and DTI [3]. This apparent contradiction can be resolved if considering that seat interface pressures are not a reliable predictor for deep tissue loads [15]. The present CII is inherently designed to predict deep tissue loads, and although the small sample size in this study did not reflect weight differences between individuals with SCI and controls, it is evident from the theory of the CII that obesity increases deep tissue loading (and consequently, the CII; Eq. (8)). This agrees very well with clinical reports of higher prevalence of DTI in obese patients [3], which provides additional support for the utility of the CII.

The theory of indentation of an elastic layer with finite thickness by a rigid half-sphere was adopted from contact mechanics and used herein to depict the concept of influence of the anatomical/geometrical
parameters relevant to DTI in the buttocks, that is, $R$ and $t$, and the $BW$, on the loading state in the muscle. It was demonstrated that theoretically, $R$ and $t$ have a similar extent of effect on peak muscle loading, and that a combination of thin muscle and sharp IT generates increased IT-muscle contact pressures, which are further pronounced when $BW$ increases (Eq. (8)). Additionally, it should be noted that scarring from a previous ulcer, initiation of a new ulcer and edema can each increase the muscle stiffness ($E$) in paraplegic and quadriplegic patients [12], which, according to Eq. (6), will similarly increase peak muscle loading in proportion with $\sqrt{E}$. Hence, other than $R$, $t$, and $BW$, muscle tissue stiffness $E$ may also be considered in a biomechanical risk assessment, particularly for patients who suffered pressure ulcers in the past: $R$, and $t$ can be measured in the clinic using conventional ultrasound [21], whereas $E$ may be measured with ultrasound-elastography, using the same hardware [16].

In summary, for the purpose of biomechanical risk assessment in patients susceptible to DTI, this communication suggests a new anatomical index, the Compression Intensity Index: $CII = (BW/Rt)^{1/2}$, which depends on the body weight ($BW$), radius of curvature of the ischial tuberosities ($R$) and thickness of the underlying gluteus muscles ($t$). Following measurements of $R$ and $t$ using MRI or ultrasound, CII can be used as a practical, quick, economic approximation of the loading intensity in muscle, and hence, can be integrated into risk assessment of DTI which is now being discussed intensively in the NPUAP and EPUAP meetings. Since there is no need to perform highly-specialized biomechanical numerical analyses for estimation of muscle loading state by means of the CII, the present technique may be particularly useful, pending larger-scale clinical studies (employing ultrasound), for implementation in small or remote medical facilities, including nursing homes and community clinics.

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References


