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A Finite-Element Biomechanical Model for Evaluating Buttock Tissue Loads in Seated Individuals with Spinal Cord Injury

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Abstract. Pressure ulcer remains to be a serious problem in today's healthcare, which has a detrimental impact on the quality of life of the wheelchair users, especially for those with spinal cord injury, and imposes a tremendous economic burden on health care. Development of effective prevention strategies and treatment modalities of pressure ulcer calls for improved understanding of soft tissue response to external loading. Initial evidences suggest that the stress in deep tissue surrounding a bony prominence may reach multiple times higher than that at the superficial skin, however, interface pressure still is the only available clinical tool to assess sitting load, which provides little to no insight to the situation in deep tissues. Moreover, experimental evaluation of the load transfer, from superficial through multiple layers of soft tissues to the load bearing bones, remains nearly impossible. Therefore, finite element models are a powerful tool to help examine biological structures and predict outcomes from potential interventions. However, to date, successful finite element models for simulating buttock tissue response in sitting are rare. In this chapter, the authors present a comprehensive description of the methodology building and validating a 3-dimensional finite element model for human buttocks with the consideration of the joint configuration and realistic boundary conditions in a sitting posture. In addition, with the validated model, sitting induced soft tissue deformation, internal pressure, and von-Mises stress were computed in a practical application evaluating a novel seating system designed to prevent pressure ulcer formation and promote healing of existing pressure ulcer. Along with the methodology of establishing and validating the buttock finite element model for sitting research, and the application example, the authors provide explicit opinions for various weaknesses usually seen in this field and suggest possible future directions for deepening our knowledge and advancing our technologies for performing successful and practical finite element simulation for buttock tissue.

1 Introduction

1.1 Unsolved Questions and Gaps in Pressure Ulcer Research Call for Mathematical Simulation

An improved understanding of soft tissue response to external loading such as in sitting may afford improved pressure ulcer (PU) prevention and treatment

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strategies. Although there are concerns and initial evidences that the local tissue stress in deep tissue layers may have been 3~9 times higher than that at the superficial skin (Bouten et al. 2003; Linder-Ganz et al. 2004; Sun et al. 2005), currently, interface pressure still remains the only available clinical tool to assess sitting load, which provides little to no insight to the situation in deep tissues. Moreover, a full-field experimental evaluation of the load transfer, from the superficial layer to the load bearing bones through multiple layers of soft tissues, remains nearly impossible. It is, therefore, anticipated that those difficulties associated with experimental measurements can be overcome by computational modeling.

1.2 Why Finite Element Modeling in PU Research

Mathematical finite element (FE) models, which allow integration of anatomy, physiology, and clinical observations, can be a powerful tool to help examine biological structures (Mow et al. 1984; Bradley et al. 1997; Keyak et al. 1998; May-Newman et al. 1998; Ferrant et al. 1999; Hagemann et al. 1999; Miga et al. 1999; Sonka et al. 2000) and predict outcomes from potential interventions (Bro-Nielsen et al. 1996; Bielser et al. 1999; Mazella et al. 1999). To analyze soft tissue response for a complex anatomical structure and address non-linear material behavior, it is indispensable to use computer FE modeling.

An improved understanding of the response of deep muscle tissue to external compressive load may afford improved prevention and treatment strategies. However, an experimental evaluation of such response on human remains impossible. Therefore, related studies rely on computational modeling (Todd et al. 1994; Ragan et al. 2002; Linder-Ganz et al. 2004; Sun et al. 2005; Lin et al. 2006) and animal models (Salcido et al. 1994; Dupont-Versteegden et al. 1998; Bosboom et al. 2001; Bosboom et al. 2003; Gefen et al. 2005).

1.3 History of FE Modeling Related to PU Research

There have been several attempts in using FE analysis to evaluate sitting induced mechanical response in deep tissue of the buttocks (Chow et al. 1978; Dabnichki et al. 1994; Todd et al. 1994; Brosh et al. 2000; Ragan et al. 2002; Oomens et al. 2003; Lin et al. 2004; Kuroda et al. 2005; Linder-Ganz et al. 2007; Linder-Ganz et al. 2009). Chow and Odell (Chow et al. 1978), using a simple axi-symmetric buttock FE model, found that the buttock tissue experienced small deformation when using a floating support design cushion. Dabnichki et al. (Dabnichki et al. 1994), employing contact elements in a simple 2D FE buttock model, found that the maximal compressive stress and displacement generated close to the bony prominences depended on the surface conditions, with the rough compliant surface generating the least compressive stress and tissue deformation. Oomens et al. (Oomens et al. 2003), using a simple 2D FE model, also studied the mechanical responses of the deep tissues related to cushion properties. They found that using a soft foam layer on the cushion provided a reduction in the normal stresses at the interface and in the deep tissue as the result of an increase in contact areas. These studies suggested an optimal method to reduce interface pressure and stress/strain on the skin and in the deep tissue of the buttock.

However, several intrinsic weaknesses are associated with the majority of the above models.

1.3.1 Limitations of Previous FE Models

Problems with geometry of current FE models

First, some of the models used simple geometrical objects to approximate the complicated anatomical structure (Brosh et al. 2000; Oomens et al. 2003; Kuroda et al. 2005). Second, for those models based on MRI recording, the MRI images were **not** performed in a setup close to real sitting, but in a supine posture (Protz et al. 1990; Todd et al. 1994) which differs intuitively from that in sitting posture in joint configuration and soft tissue stress-strain conditions.

It is well known that the accurate FE simulation of biomechanics needs accurate 3D geometry and realistic material models of a biological structure. Therefore, it is necessary to develop a comprehensive methodology for using 3D FE modeling technology in evaluating soft tissue response to external sitting load.

Problems with material models

Biological soft tissues have viscoelastic, anisotropic, and incompressible properties (Kosiak 1961). The description for such a complicated behavior of the soft tissues may be the bottle neck of applying the FE modeling to simulate clinical problems. Most currently reported models use material properties available in the literature and used the material properties in a way which is substantially different from the real scenario in the buttocks during sitting, in the following aspects.

In sitting, the soft tissue, especially the skeletal muscle, is subjected to ***compressive*** load ***transversely*** about the fiber orientation, while material properties given in literature are mostly obtained ***longitudinally*** along the fiber orientation, and mostly under ***tension***. Our laboratory data showed that, for the same tissue sample, material properties can be substantially different between testing in longitudinal (parallel to fiber direction) and transverse directions (Figure 1).

Those material properties used were from literature, mostly, obtained from fresh frozen cadaver or animal tissue without measuring tissue perfusion (Linder-Ganz et al. 2004), which inevitably introduced unknown discrepancies. Published values of Young's modulus of component tissue vary by up to an order of magnitude, presumably due to the method of measurement or estimation (Mason 1977; Agache et al. 1980; Larabee 1986; Larabee et al. 1986; Sarvazyan et al. 1994; Azar et al. 2000; Samani et al. 2001; Schnabel et al. 2001). Evidence shows the material property of biological tissue change substantially with time after dissection and with method of preservation (Metz et al. 1970). Literature has reported that material properties for postmortem soft tissues are significantly different from those of living tissues(Fung 1993). The material property of skeletal muscle is affected by tissue perfusion and muscle tone, i.e. a physiologically low level muscle contraction seen in skeletal muscle (Masi et al. 2008). Therefore, *in* *vitro* material test uses tissue samples which are excised from innervation and blood supply which hardly an appropriate "sample" for the living tissues.

Most earlier FE buttock models for sitting used linear material models to describe the non-linear biological tissue, which added more uncertainties on the FE model's output. More frequently, isotropic characteristic was assigned to the material models.

Another major weakness of current FE models for buttocks was that most of these models used a single material property for all soft tissue (Chow et al. 1978; Dabnichki et al. 1994; Todd et al. 1994) without differentiation among muscle, fat and skin. Also from Figure 1, it is evident that these tissues do not possess the same or even similar material properties.

No FE buttock model so far considered the specific material properties for the soft tissues in individuals with spinal cord injury (SCI) or pressure injury. Pressure or other mechanical loads cause alterations or adaptations of tissues (Sanders et al. 1995). Mechanical stress induced adaptations have been observed both in human

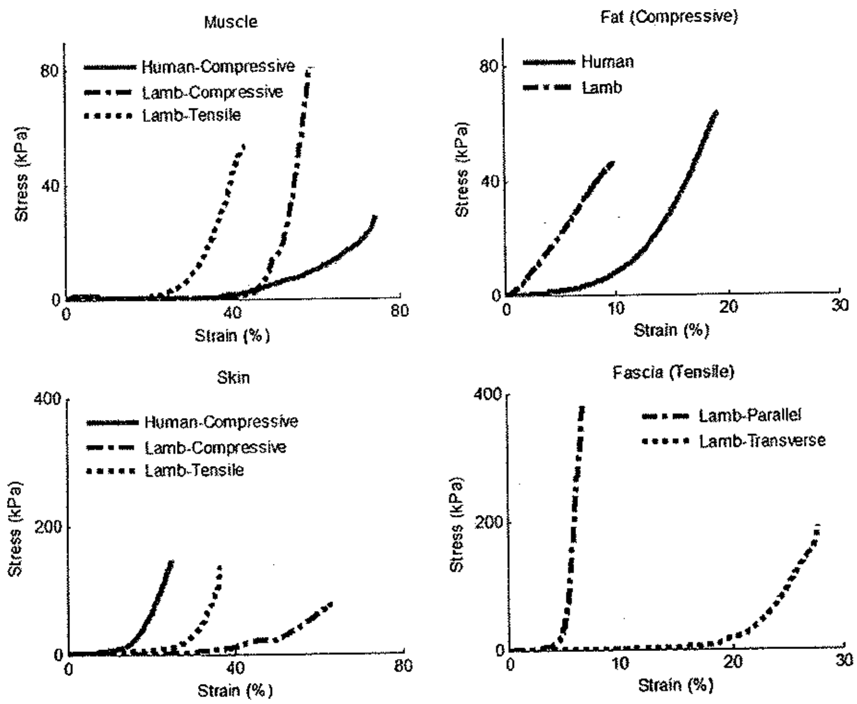


Fig. 1. Stress-strain results for muscle, fat, skin and fascia from tensile and compressive material test from human and animal tissue samples. Some of the samples were also tested in longitudinal (parallel to fiber orientation in figure) and transverse directions.

(Grimby et al. 1976; Stilwill et al. 1977; Scelsi et al. 1982; Lotta et al. 1991; Martin et al. 1992; Greve et al. 1993; Round et al. 1993; Edsberg et al. 1999; Edsberg 2001) and in animal models (Lieber et al. 1986a; Lieber et al. 1986b; Roy et al. 1991; Goldstein et al. 1998; Castro et al. 1999; Castro et al. 2000; Sanders et al. 2001; Wang et al. 2003; Landry et al. 2004) with and without SCI. Tissue at or adjacent to the PUs have been found to undergo significant adaptation or remodeling as a result of sustained pressure (Edsberg et al. 2000). Our preliminary data on rat tibialis anterior (TA) muscle stiffness also demonstrated such substantial difference between pressure injured and intact tissues. Figure 2 shows the plot of load-deformation curves for samples from intact TA and pressure-injured TA from the same rat. For obtaining these data, the TA on one side was kept intact while the other side was subjected to 200kPa compression for 4 hours 24 hours before the load-deformation testing.

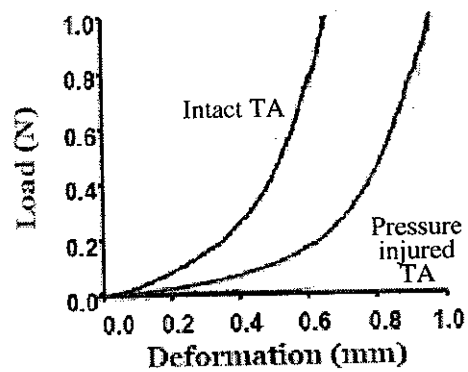
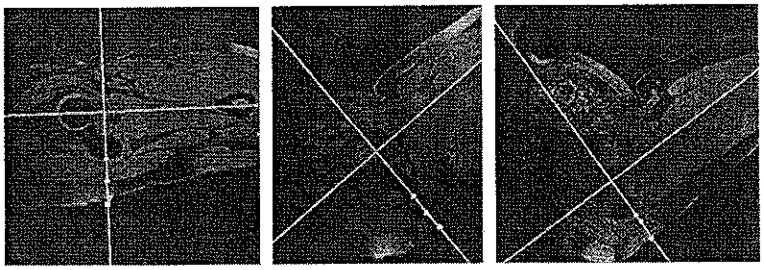


Fig. 2. Load-deformation data for *in-vivo* tests on rat tibialis anterior (TA) muscle. Data came from the pressure-injured TA and its contralateral intact TA on the same rat.

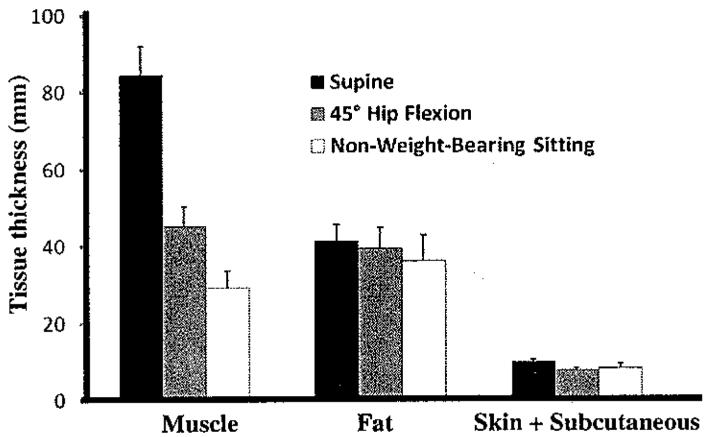
2 Methodology on Building an FE Model for Buttocks for a Seated Individual

2.1 The Importance of Building the Model in a Seated Configuration

As emphasized previously, the FE model for buttocks for a sitting posture should be built upon the geometry obtained from a sitting posture which is intuitively different in anatomical configuration (joint and tissues) from that of a supine or standing posture. In sitting, the hip and knee joints are both flexed to close to 90°, therefore, the loading relationship between segments is different between these 2 postures. With the flexion of the hip joint, the gluteus muscle slides laterally away from the ischial tuberosity (IT) (Kaplan 2003). Therefore, the tissue being compressed in sitting by the ITs is not that seen distally to IT in full hip extension as in standing or supine posture. Also because of this lateral sliding of the gluteus muscle, in sitting, the IT has less padding tissue to resist the pressure load, which can be mistakenly interpreted by taking the anatomical geometry in a supine posture. Our data of tissue thickness from 11 subjects at different hip flexion angle



A. Buttock tissue thickness measurement on MRI images at different hip flexion angles



B. Data of measurement from A.

Fig. 3. Soft tissue thickness under the ischial tuberosity in three different hip flexion angles (0°, 45°, and 90°). Data were from MRI images of 11 able-bodied subjects.

measured from MRI images (Figure 3A) clearly showed this substantial difference in tissue composition and thickness between sitting and supine postures.

Therefore, the geometry of a FE model for buttocks for sitting related research should be acquired in a sitting posture or a posture resembling the hip and knee joint configuration of that of a sitting posture.

2.2 Building the Geometry of the Model

2.2.1 Building the Geometry in a Simulated Sitting Position

In a regular MRI scanner, an upright sitting posture cannot be achieved for a human adult because of the limited room. To overcome the restriction, we used a simulated sitting posture for acquiring the MRI images for buttock structure.

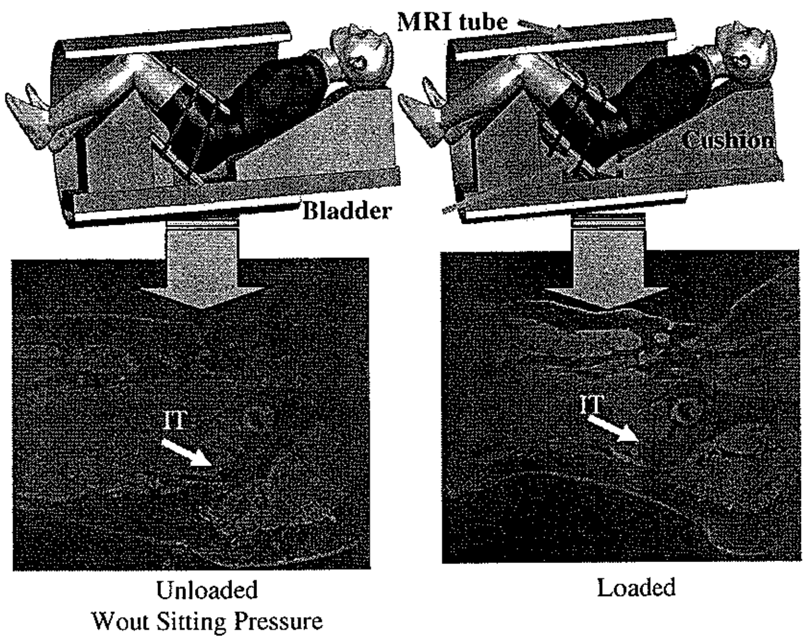


Fig. 4. Setup for recording MRI images from buttock-thigh area in a simulated sitting posture with simulated sitting load applied. Upper row: MRI setup to measure buttock-thigh structure under two loading configuration (Left: Unloaded: Without Sitting Pressure; Right: Loaded: With Sitting Pressure) for the simulated sitting posture. Two belts tied the cushion with the buttocks. An air bladder was placed in between of the 2 layers of the cushion. Inflation and deflation of the air bladder provided different loading levels. Lower row: Corresponding MRI images.

A custom built apparatus made of plastic foam and rubber was used to place the subject in a simulated sitting posture (Figure 4). Two loading conditions were used (Unloaded: without sitting pressure; Loaded: with sitting pressure) during the MRI recording. The sitting pressure was applied to the buttocks with an adjustable cushion placed under the buttock-thigh. The cushion was a sandwich structure consisting of two layers of stiff materials with a rectangular air bladder in between, while two belts tied the cushion tightly against the buttocks. Inflation and deflation of the air bladder provided different loading conditions as shown in Figure 4. The magnitude of the applied sitting pressure for the "With Sitting Pressure" configuration was selected from an actual average interface pressure under buttock (152.6±20.9mmHg, 20.3±2.8KPa) measured with a pressure mapping system, X2 (Xsensor Technology, Calgary, Canada) for this subject in an upright sitting posture. MRI images obtained from "Without Sitting Pressure" were used for development of the geometry of the FE model and those from "With Sitting Pressure" were used for validation of the FE analysis.

MRI images of buttock-thigh were obtained (Figure 4) for the above two loading conditions in one simulated sitting posture (80° flexion for hip and 90°

flexion for knee). The MRI parameters were 1.5 Tesla, 576x576 matrix, 35x35 cm DFOV, 0.6 mm inter-slice thickness.

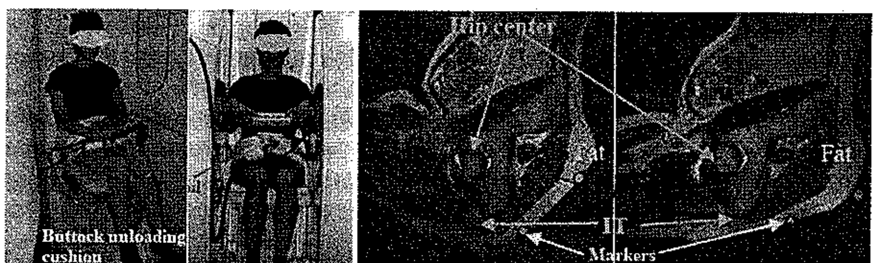
2.2.2 Building the Geometry in a True Sitting Position

Although the above simulated sitting posture works well for acquiring the buttock geometry in a joint configuration similar to sitting, there are drawbacks. First, the simulated posture can only be achieved for individuals with relatively smaller body build, based on our experience, not taller than170cm in height. Second, since the subject is away from a load-bearing sitting posture, the sitting load can only be applied to the buttocks from the bottom, which is not the natural loading pattern during sitting. Therefore, using a setup which allows a true sitting posture be scanned provides an appropriate alternative.

In our study (Figure 5) and the work from Linder-Ganz (Linder-Ganz et al. 2009), an open MRI setup has been used for scanning the buttocks in a true sitting posture. In our study, a regular seat cushion without metal parts was used between the subject's buttocks and the seat. Additional cushioned material prevented the direct contact of the subject's body with the coil. MRI images were acquired in 2 conditions, "without buttock load" and "with buttock load" when the subject was in the upright sitting posture.

For "without buttock load" condition, an enhanced lumbar support was used at L4-L5 to support the spine. A cushion (buttock unloading cushion) was placed underneath the thighs to bear the sitting load while left the buttocks load free (Figure 5A, left). For the "with buttock load" condition, the subject remained in the above posture but the buttock unloading cushion was removed to allow the subject sit directly on the seat cushion with the upper body weight born by the buttocks. The buttock-cushion interface load was measured by using XSensor (XSensor Technology, Calgary, Canada) while the seating system was out of the MRI.

Volumetric MRI images of the right buttock-thigh including coccyx (256x256 matrix, 35cm FOV, 288 slices, 0.5mm thickness) were obtained using the 0.6T FONAR scanner. Several MRI opaque markers were placed on buttock skin for the purpose of registration of the images for later processing (Figure 5B).



Without buttock load With buttock load
A. Stand-Up MRI setup

Without buttock load With buttock load
B. MRI images

Fig. 5. Open (Stand-Up) MRI (Upright™, FONAR Corp., Melville, NY) setup to record buttock-thigh structure under sitting load. MRI images corresponding with and without buttock sitting load are shown.

2.3 Mesh Generation of the Final Geometry

After the imaging, the FE model geometry is usually created based on the 3D reconstruction of the MRI images under the non-weight bearing configuration. Contours of the femur, pelvis, skin, fat and muscle are identified and outlined from these images by using segmentation software such as WinSurf (SURFdriver, Kailua, HI).

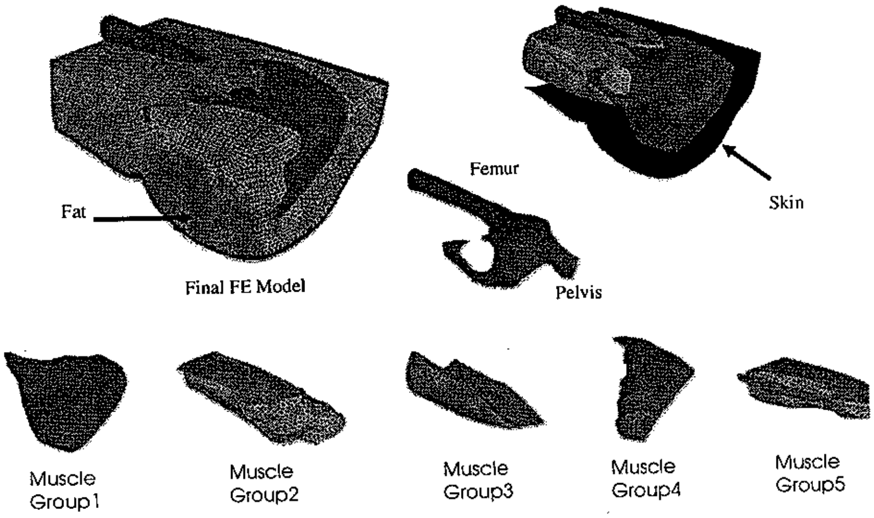


Fig. 6. Structural elements of the FE buttock model

Muscle is one of the primary tissues in buttock structure and is thought to be the most susceptible to pressure induced injury, especially in the vicinity of bony prominence such as ischial tuberosity where the pressure ulcers usually form. Therefore, the FE simulation for buttocks in sitting usually focuses on the stress/strain within the muscle tissue. The massive volume, separated locations, and various orientations of the buttock muscles bring issues for creating the muscle as one object in a buttock FE model. Our current model uses five groups to represent the various location and orientation of buttock muscles. Selection for muscles in each group depends on the location, and the line of action of each muscle. Considering the line of action for each muscle group is for applying initial muscle tone to each group, which will be described later. However, we believe that grouping muscles or not depends largely on particular application and the focus for outcome evaluation. The five muscle groups used in our model are:

Group1: *Minimus, Medius and Maximus Gluteus.*
Group2: *Adductor Longus, Adductor Brevis, Adductor Magnus, and Pectineus.*
Group3: *Biceps Femoris (long head), Semitendinosus, and Gracilis.*

Group4: *Piriformis, Superior and inferior Gemellus and Obturator Internus.*
Group5: *Vastus Lateralis, Vastus Intermedius, Vastus Medialis and Rectus Femoris.*

The segment information is then translated into HyperMesh (Altair Engineering, Inc., Troy, MI) to create the FE mesh. The final FE model consists of 453,502 four-node tetrahedral solid elements for the pelvis, femur, inner side of the skin, fat and five muscle groups, and 33,924 three-node triangle membrane elements for the outer side of the skin (Figure 6).

2.4 Selecting Material Model for Buttock FE Models

As pointed out previously, the biological tissue is non-linear, anisotropic, nearly incompressible material with viscoelasticity. Moreover, soft tissues in buttocks experience deformation as large as about 70% in practical sitting (Lin et al. 2004). Therefore, material model for FE simulation for buttocks in sitting should be selected accordingly. Adding to this complexity is the significantly altered material property for denervated tissues below the level of injury in SCI individuals. *In vivo* (blood perfusion preserved) compressive creep test on tibialis anterior (TA) muscle flaps from 10 neurologically intact and 8 chronic SCI rats in our laboratory revealed a 44.9% increase in SCI group in compressive deformation after 500s holding of a constant load (the constant holding load was chosen as 30% of the load needed to achieve 30% deformation for each TA muscle). This finding suggested that, in addition to the rapid muscle atrophy (Scelsi et al. 1982; Lotta et al. 1991), muscle tissue alters its fundamental properties post SCI and these alterations lower their pressure endurance threshold significantly (Daniel et al. 1981). This phenomenon calls for its proper representation in the material model chosen for simulation sitting for SCI population. Unfortunately, research on the changes of tissue material properties post SCI is scarce to none. Currently FE models, even when the focus is for evaluating sitting load in SCI wheelchair users, still employ material properties from the literature obtained from unrelated tissue samples.

In our developed FE model, static equilibrium equations with large deformation were assumed in the material model since pressure sores develop after prolonged loading with large deformation of the soft-tissue. The elastic *Moony-Rivlin* model (Dabnichki et al. 1994) (1^{st} order polynomial model), which can account for large deformation behaviors of materials, was employed for skin, fat and muscle. The model was based on the following strain energy function W (Eq. 1):

$$W = C_{10}(\bar{I}_1 - 3) + C_{01}(\bar{I}_1 - 3) + \frac{1}{D_1}(J-1)^2, \qquad (1)$$

W is the strain energy per unit of reference volume; J is the total volume ratio; C_{10}, C_{01}, and D_1 are temperature-dependent material parameters; $\bar{I}_1$ and $\bar{I}_2$ are the first and second deviatoric strain invariants defined as following equations (Eqs. 2 and 3):

$$\bar{I}_1 = \bar{\lambda}_1^2 + \bar{\lambda}_2^2 + \bar{\lambda}_3^2, \tag{2}$$

$$\bar{I}_2 = \bar{\lambda}_1^{(-2)} + \bar{\lambda}_2^{(-2)} + \bar{\lambda}_3^{(-2)}, \tag{3}$$

The deviatoric stretches $\bar{\lambda}_i = J^{-1/3}\lambda_i$; Here, λ_i are the principal stretches.

The material parameters for skin, fat and muscle were determined based on values obtained from literatures (Dabnichki et al. 1994; Oomens et al. 2003). For all soft-tissues, D_1 was determined based on the method reported by Dabnichki et al, (Dabnichki et al. 1994) with an assumption that the materials were nearly incompressible (*Poisson ratio* $\cong$ *0.485*).

2.5 Assigning Boundary Conditions for Buttock FE Models

The boundary conditions for buttock FE models in sitting reflect the anatomical and practical scenarios in human sitting. In sitting, soft tissues are compressed by the sitting load coming from the seat-skin interface. The induced movement of the tissues is among the interests in sitting simulation. Movement does not usually occur in proximal-distal due to the anatomical restrain. Also the tissues on anterior side of the pelvic and femur do not usually move to anterior. To achieve efficient simulation, usually one half of the buttocks is used for FE simulation. The other half is assumed to experience the same loading condition with the same response.

Based on the above description, three boundary conditions (*BC*) are employed in our developed FE model (Figure 7). The femur and the pelvis are assumed as rigid bodies and constrained to obviate rigid body motion (BC1). The medial plane was constrained against the medial-lateral motions because of the symmetric

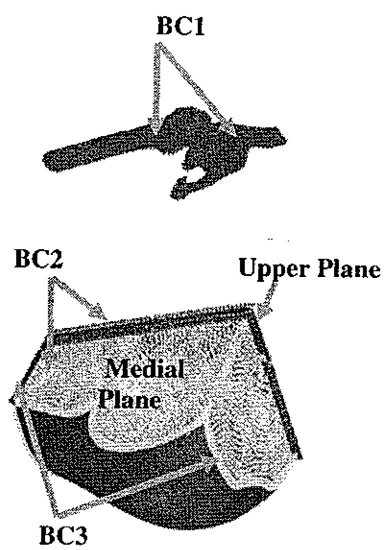


Fig. 7. Boundary conditions applied in the FE model

condition of the buttocks (BC2). Also for BC2 was that the upper plane was constrained from the anterior-posterior movement. The ends of tissues that connect to the rest of the body (distal end to the thigh, proximal end to lumbar region) were constrained against longitudinal motions (BC3).

2.6 Assigning General Loading Conditions for Buttock FE Models

When load from the sitting was considered as the specific loading applied as the input to the FE model, two general loading conditions (GLC in Figure 8) should also be included in the model, based on physiological facts.

First, in a sitting posture, buttock skin is not in a relaxed condition but with a low-level tension induced by hip flexion, which is represented by using an initial strain on the skin layer of the model (GLC1) (Figure 8). This initial strain is defined as the strain of the buttock skin when the subject moved from a full-hip extension posture to an upright sitting posture, when the hip and the knee both flex at about 90°. The value of this strain used in our current model was obtained from a single subject by calculating Green strain (Bonet et al. 1997) of painted dots on subject's buttock skin. Photographs were taken at 0° hip/knee flexion and again at a hip flexion angle of 80° and knee flexion of 90°. On these photographs, the array of dots painted on the buttock skin was digitized using custom codes on Matlab platform (Figure 9). Green strain was then calculated for an area of $15 \times 15 mm^2$ as 43.6±10.4% and 11.2±0.7% for the axial and transverse directions, respectively.

Secondly, skeletal muscles are consistently under a tension from a low level contraction, resting tone (Masi et al. 2008) or resting tension. To simulate this resting tone, we apply 1% of maximal muscle force estimated from $F_{max} = \sigma_{max} \times$ *Physiological Cross-Sectional Area (PCSA)*, to each group of muscles along the line of action. In the above equation, σ_{max} is the maximal muscle tension (Morrey 1990) (GLC2 in Figure 8).

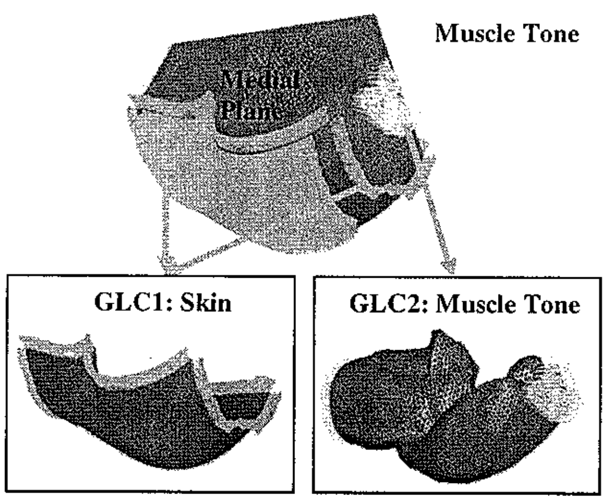


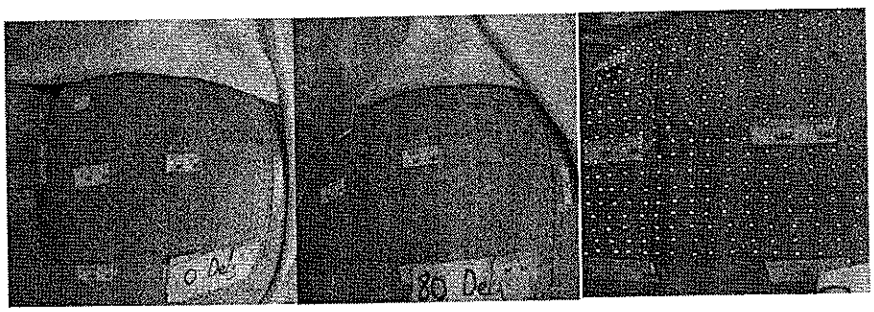
Fig. 8. General loading conditions applied in this FE model

3 Results

3.1 FE Model Solution

Upon establishment of the geometry and material model, the FE model is ready for a simulation. To solve the model simulation, the sitting interface pressure is used as the model load input. For this purpose, a sitting contact area is precisely identified from the corresponding MRI images and the sitting load is applied to this area.

Buttock contact pressure corresponding to the MRI scanning in loaded condition is used. In model development and validation stage, to be consistent with the loaded condition shown in MRI images, the FE model should be fed with the same input load to be able to perform the comparison in the validation process. Thus, we used a uniform contact pressure of 20.34kPa (152.60mmHg) because a constant pressure was actually applied to the buttocks when MRI images of loaded condition were taken. With this load input, the FE model was solved using ABAQUS software (ABAQUS 6.5, ABAQUS, Inc., Providence, RI).



A) 0° hip flexion B) 80° hip flexion C) Digitized dot array

Fig. 9. Measurement of buttock skin initial tensile strain induced by a sitting posture relative to a standing posture

3.2 Outputs of the FE Model Simulation

I. Soft tissue deformation under sitting load

This prediction is performed for a sagittal piece of the entire buttock-thigh. This sagittal piece had a thickness of 12mm with its medial-lateral center at the tip of the ischial tuberosity. In this sagittal piece, the gross deformation of each layer was calculated using the changes in the coordinates of the FE nodes, 42 nodes per layer, in anterior-posterior direction when the buttock-thigh was loaded by 20.34 kPa.

II. Soft tissue internal stress induced by the sitting load

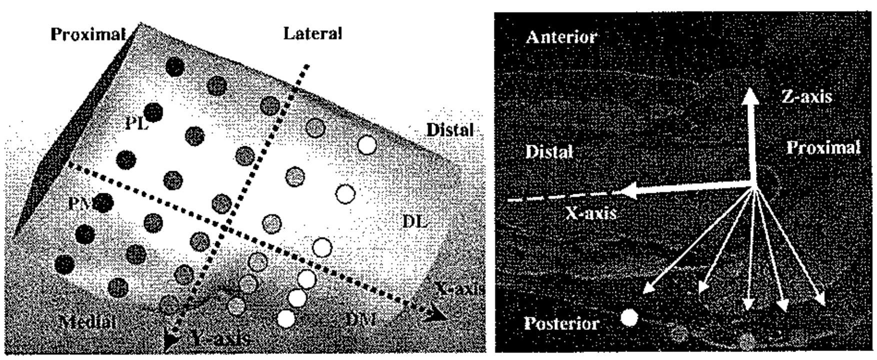
From the FE analysis, the internal pressure distribution and von-Mises stress distribution can be obtained for the entire or selected part of the structure. In our model, we calculated the internal stress for the muscle, fat and skin for the entire buttock-thigh, particularly in the area below the ischial tuberosity. Although it still is uncertain what mechanical parameters are the most relevant to tissue damage (Oomens et al. 2003), evidences supported that living cells are more vulnerable to deformation than to a high hydrostatic pressure (Bouten et al. 2003). Therefore, the von-Mises stress, which is related to the deformational energy stored in the material (Todd et al. 1994; Oomens et al. 2003; Linder-Ganz et al. 2004), was analyzed to identify the risk of possible tissue damage by sitting load.

3.3 Methodology on Validating an FE Buttock Model for a Sitting Posture

A FE model is a mathematic model which is built upon mathematical description of the structure and the materials. For a model to have the practical value in estimating the interested outcomes from successful simulation of the mechanical environment within the structure represented by the model, it should be validated before putting into practical application. An un-validated model, however complicated, can server at the best a qualitative tool. For a mathematical model simulating physiological process within a human body, it is always a challenge when it comes to the time for validation. Therefore, quite a few FE buttock models remained un-validated or were only validated using parameters which were used as the model inputs such as the interface pressure on the seat-buttock interface.

There are two ways to validate a FE model, through experiment on a physical model (Linder-Ganz et al. 2009), and through imaging measurement from a real sitting. Our developed model was validated using the imaging method.

Upon solved, the FE model was validated quantitatively via comparison of the output of the FE simulation with measurements from the MRI images. Two comparisons were performed: 1) the sitting induced gross displacement of soft tissues; 2) the sitting induced position shift of the muscle underneath the ischial tuberosity (Group1).



A) A posterior view of the coordinate system and the ROIs.

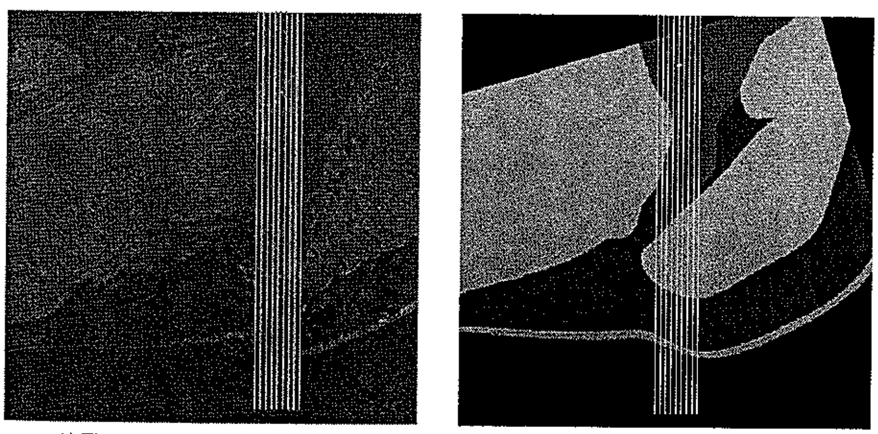
B) A sagittal MRI image showing the coordinate system and the ROIs.

Fig. 10. Validating FE model by comparing the imaging measurements and the simulation results of selected regions of interest (ROI)

I) Gross validation. *Sitting Induced Gross Displacement of Soft Tissues Computed by FE Analysis and that Measured from MRI Images:* It was assumed that the bony structure remained the same when loaded. A Cartesian coordinate system was defined based on the femur-pelvis bony structure. This coordinate system took the center of the femoral head as its origin. The X-axis was pointing to the distal along the femoral shaft. The Y- and Z- axes were pointing to the medial and the superior, respectively (Figure 10). Thirty regions of interest (ROIs) were identified over the skin of the sitting area (Figure 10). For measurement on MRI images, the morphological alteration of buttock soft tissues induced by sitting pressure of 20.34±2.79KPa was identified for these ROIs by comparing MRI images obtained in loaded condition to the matching image from unloaded condition. On each image, vectors pointing from the origin of the coordinate system to the specific locations on the skin were constructed to compute the gross displacement. In order to compare the differences in different portions of the buttock, 4 regions, the distal-medial (DM), distal-lateral (DL), proximal-medial (PM) and proximal-lateral (PL) regions, were defined (Figure. 5).

The same coordinate system mentioned above was established for the FE model. The 30 ROIs were also identified for the FE model for both the unloaded "Without Sitting Pressure" and loaded "With Sitting Pressure" configurations. The sitting induced changes of the coordinates of the FE nodes in these ROIs were taken as the FE predicted gross displacement. These displacements were then compared with the measurements from MRI images.

II) Selected validation for more details regarding the simulation focus. *Position Shift of the Muscle Group1 beneath the Ischial Tuberosity:* In the sitting condition, it is apparent that the tissue underneath the ischial tuberosity is loaded more than tissues at other locations. Therefore, the muscle here may be pushed away from its



A) The selected area on MRI images for calculating muscle Group1 shift

B) The selected area on the FE model for calculating muscle Group1 shift

Fig. 11. Validating FE model by comparing the imaging measurements and the simulation results of a selected region underneath ischial tuberosity

original position. This shift of position of muscle Group1 can be used as a validation parameter. A small area right beneath the ischial tuberosity was selected to calculate the shift of the muscle Group1 induced by sitting load of 20.34 kPa. Taking the tip of the ischial tuberosity as the center, this area covered ±15mm in the proximal-distal direction and ±6mm in the medial-lateral direction (Figure 11). For this volume, in the medial-lateral direction, 12 sagittal slices were taken for the FE/MRI models. On each of these sagittal slices, 11 lines in anterior-posterior direction were defined with a distance of 3mm between neighboring lines. Therefore, altogether 132 anterior-posterior lines were determined in this volume. For MRI model, the proximal-distal shift of this volume under sitting load was measured as the average proximal-distal shift of these 132 lines. For the FE model, the proximal-distal shift of this volume was obtained by identifying the changes of the coordinates of the FE nodes within this volume.

3.4 Application of FE Models for Evaluating Buttock Loading in Sitting

Once validated, the FE model can be used to simulate various practical scenarios. Based on the purpose of the application, appropriate outputs can be selected. Most frequently, buttock FE models for sitting are used for evaluating different seat designs, including shape or material of the cushion.

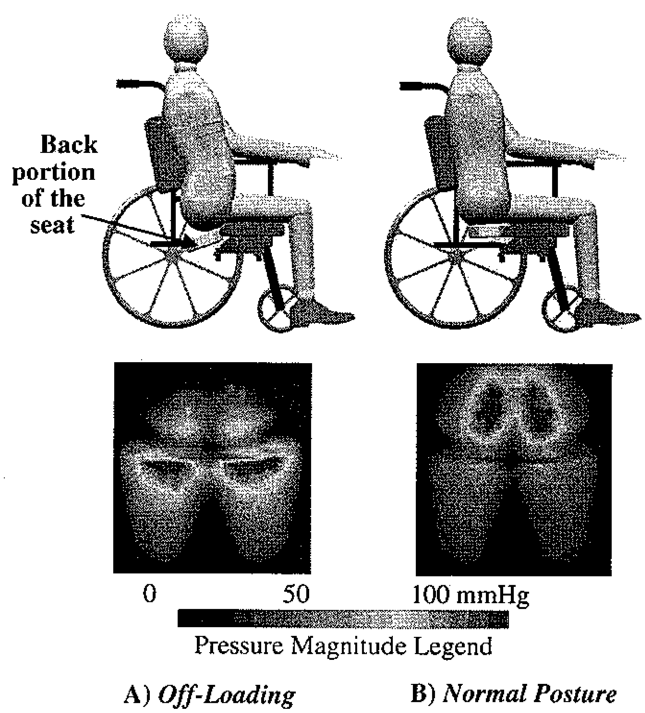


Fig. 12. Two sitting configurations of the studied seat

The above described FE model (Makhsous et al. 2007a) was used in an application to evaluate a novel seat cushion for its effectiveness on reducing the internal stress around the ischial tuberosity. The seat design features 2 configurations, *Normal* and *Off-Loading* (Makhsous et al. 2003; Makhsous et al. 2007b). A *Normal* posture was described as sitting upright on a flat seat and with a flat backrest, while an *Off-Loading* posture was sitting upright on a seat with partially removed ischial support, and with an enhanced lumber support (Figure 12). The Off-Loading configuration was intended to release the ischial load from sitting.

For evaluating this seat, the FE model input was the averaged interface pressure map measured from 35 able-bodied individuals (41.3±12.1years; 72.1±12.6kg; 168.0±8.5cm) in Normal and Off-Loading configurations (Figure 12). In the *Normal* configuration, the interface pressure was concentrated within the vicinity of ischial tuberosities, with the thighs taking substantially less interface pressure. In the *Off-Loading* posture, the concentrated interface pressure at ischial tuberosities was mostly shifted towards the thighs, and the middle part of the thighs took the most loads.

With the interface pressure as the load input to the FE model, simulation provided estimation of the internal stress and strain within buttock structure shown in Figure 13 and Figure 14.

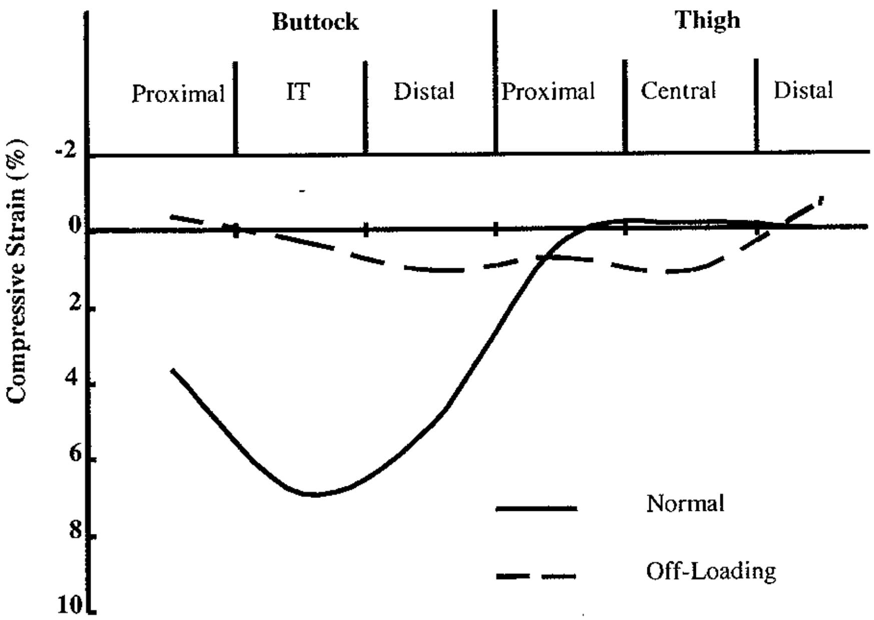


Fig. 13. Anterior-posterior compressive strain (%) of the muscle layer in sagittal plane in a portion covered 15mm thickness buttock-thigh from 7.5mm medial to the ischial tuberosity (IT) to 7.5mm lateral to IT

Compressive Strain on Muscles (Figure 13). The larger compressive strain in the *Normal* sitting (6.9%), which was seen underneath the ischial tuberosity, was shifted toward the thigh with a largely decreased value (0.3%) in the *Off-Loading* condition. On the other hand, the compressive strain on the center of the thigh region was increased from 0.2% in the *Normal* condition to 1.1±0.6% in *Off-Loading* condition.

Compressive Stress and von Mises Stress Distribution (Figure 14). The predicted compressive stress and von Mises stress distributions in the entire buttock-thigh structure and in muscles below the IT are shown in Figure 14. It was seen that, in *Normal* configuration, the high-pressure and stress region was at the location right beneath the bony prominence of ischial tuberosity. However, this high-stress region was relocated to the thighs with a reduction of magnitude in the *Off-Loading* posture (Figure 14).

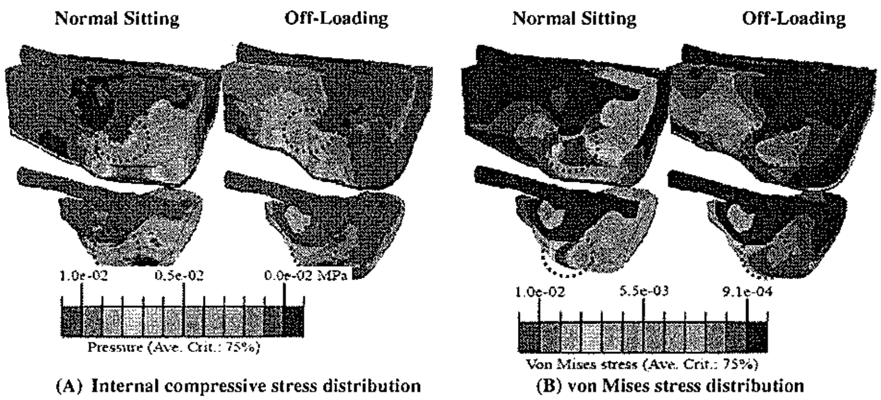


Fig. 14. FE simulation results on stress for a split seat in 2 of its configurations, *Normal* and *Off-Loading*. In each subset of the figures, the upper plots show results for the entire structure, while the bottom plots show data for the muscles below the ischial tuberosity.

This simulation confirms that the pressure relief provided by the *Off-Loading* posture has profound effect in decreasing the stress ***within*** the deep tissues of the buttocks. FE simulation shows that at the same time of releasing interface pressure, the *Off-Loading* posture significantly redistributed the internal compressive stress and von Mises stress within the deep tissues of the buttocks. In the *Off-Loading* posture, the internal compressive stress which was highly concentrated in the deep tissues at the location of the ischial tuberosity in *Normal* sitting posture were repositioned to the thigh with a substantially decreased magnitude.

The most common sites for sitting related PUs are areas close to a bony prominence. At these locations, such as the ischial tuberosity, tissue necrosis was frequently seen starting from deep in the tissues and progresses towards the superficial to become an open wound (Ankrom et al. 2005). Therefore, there is a need for a tool which can estimate the internal mechanical responses in the deep tissues, such as stress and strain, to external sitting load, to provide, from a mechanical point of view, a deep insight into the relationship between the formation of deep tissue damage and the interface pressure. With more of this understanding, it is possible to establish a criterion for evaluating designs of those new sitting systems which aim at achieving optimal pressure relief in sitting area for wheelchair users. The results in the current study proved that the *Off-Loading* posture is effective in reducing the risk of PUs by decreasing large compressive strain, high internal compressive stress in the deep tissue overlying ischial tuberosity.

It has been suggested that the muscular layer is the most *vulnerable* to pressure induced tissue damage due to its high metabolic rate and the dependence on tissue perfusion (Nola et al. 1980; Daniel et al. 1981). This vulnerability puts muscle tissue in a highly susceptible situation to localized compression, eventually leading to tissue degeneration in the form of a deep pressure injury. Therefore, our findings about the concentrated stress in the deep layer of the muscle tissue may be considered the evidence for this kind of deep tissue injury.

4 Future Direction for FE Models in Pressure Ulcer Research for SCI Population

FE simulation for sitting related topics has gained significant progress in recent years and proved itself as an extraordinary tool in understanding the comprehensive response to sitting loads in complex structure within buttocks. As demonstrated in this and other chapters, FE models can now be developed for the buttock anatomy with sufficient fidelity and acceptable accuracy and has been more and more used in the area investigating sitting phenomena, evaluating seating system design, and possible providing useful suggestion to various applications.

However, there are several important issues still hinder the complete acceptance of FE buttock models as a perfect tool for sitting related research and applications, especially when it comes to the practical scenario of wheelchair seating in SCI population. We have identified several critical issues for FE simulation in wheelchair sittings for this population, possible solutions and potential future directions.

4.1 Improving Material Properties

I: Material properties for buttock tissues in SCI samples. With the advances in medical imaging, image processing, and reverse engineering technology, the methodology and tools to obtain an accurate geometrical representation for any complex anatomical structure are being developed and become more and more sophisticated. The geometry of the FE models has progressed from an assembly of simplified 2D objects to faithful 3D structure with incredible resemblance to the real anatomy. Compare to this progress, material models for buttock tissue modeling, the other critical issue in FE sitting modeling, have gained little advance, especially not for the SCI samples. As known by probably every researcher in FE modeling area, a faithful description for each material involved in the FE model dictates the performance of the simulation, more importantly, the applicability of the model simulation. Therefore, the lack of a database for tissue material properties for SCI population is currently the bottle neck of applying FE simulation to simulate clinical problems. As evidenced by literature and our research, tissues, especially soft tissues, experience tremendous morphological and mechanical adaptation post SCI. The loss of proper muscle tone in muscles below the level of SCI due to denervation or disturbed neuromotor control makes additional contribution to post-SCI evolvement of tissue properties. Therefore, future researches should emphasize this need and a comprehensive yet careful investigation in researching such changes in the properties of the soft tissue below the level of the injury of the spinal cord is a must. Currently, this research is scarce to none. Several directions are worth to explore.

II: Material properties in SCI individuals. Data for material properties in SCI samples is in urgent needs. Collecting comprehensive material properties for SCI tissue samples can be a very tedious endeavor, including testing samples in various conditions within this population, such as tissue from acute and chronic stages of SCI for both genders and all age groups, and for people with SCI plus some other comorbidity which may also affect tissue properties. For example, the prevalence of diabetes in SCI is 15%~20%, which is higher than the 10.7% in general population. Diabetes is a medical condition which is known to affect soft tissue properties and increase the incidence of ulcers. Therefore, data for tissue properties from individuals with SCI and with diabetes should be obtained for better FE simulation application for these individuals.

III: Material properties in tissues post SCI from animal models. Collecting material property data from human tissues in SCI population will inevitably encounter several obstacles which prevent us from collecting data in a well controlled condition. For example, *in vivo* sample testing is almost impossible to perform on human subjects, which limits the data collection to excised tissue samples with yet highly limited availability. In well designed animal model, the condition for SCI and selected comorbidity can be strictly controlled and reproduced in multiple animals. At the same time, *in vivo* material testing on tissues with preserved blood perfusion and innervation is feasible and has been conducted in various setups including those from ours. By comparing material properties from SCI (and/or together with selected comorbidity) animals with those from the intact/healthy animals, the relative changes of tissue properties can be obtained. Then these relative properties may be used to calibrate human tissue properties for SCI individuals from that of the able-bodied individuals.

4.2 Subject-Specific Model

This direction has been initially explored by some groups of researchers (Lin et al. 2004; Wagnac et al. 2008; Linder-Ganz et al. 2009). However, being subject-specific usually means more resources and more work would be invested. Therefore, a valid methodology for scaling a prototype or primitive model to each individual is promising; however, no such work has been done yet. Research in this direction should be given more attention to help push forward the FE simulation to be a practical tool in both laboratory and clinics.

4.3 Using Combination of MRI/Open MRI

As described previously, there are two different methods to acquire the geometry of the buttocks for establishing a FE model, imaging in a simulated sitting posture in a high resolution MRI setup, or in a true sitting posture in an open MRI setup with less spatial resolution. Advantages and drawbacks are associated with each of the methods. To overcome the drawbacks and to take advantages from both of them, a possible solution is to combine the two methods to obtain the geometry both accurately using the simulated sitting posture in a high resolution MRI machine and with true sitting load using an open MRI setup.

Combination of these 2 types of MRI scans in constructing accurate and realistic FE model geometry has been reported for patellofemoral joint in researches related to patellofemoral joint loading simulation (Besier et al. 2008). The concept presented in these researches can be readily adapted to FE modeling for buttocks in sitting. Images can be obtained from regular MRI with higher resolution in the simulated sitting posture and then another scan for a loaded true sitting condition can be performed by using an open MRI machine. These 2 sets of MRI images can be registered to each other to produce a final series of MRI images which describes the buttock geometry in a sitting posture with soft tissue deformation induced by true sitting load. In this way, the most accurate yet the most realistic model geometry can be generated.

4.4 Apply More Realistic Loading to the Model

Currently, most of the FE buttock models use pressure reading on the buttock-seat interface as the input to drive the simulation. However, we know that this is not what actually happens during sitting. In a sitting posture, upper body weight is born by the buttocks with a load in the direction of gravity is applied through the spine to the sitting anatomy. Therefore, a more realistic way to simulate sitting would be to apply the upper body weight through the connecting bony structure, i.e. the sacrum which connecting the upper body to the pelvis. If the model geometry includes a sacrum, the upper body weight can be applied to the sacrum directly. However, many FE buttock models do not have the sacral segment then the upper body load should be applied to the surface of sacroiliac joint. The body weight of the buttocks which is below the sacrum level should be applied to the model as distributed load in the gravity direction.

4.5 Develop FE Models for Clinical Use

As described previously, a validated FE buttock model can serve as a powerful tool providing detailed and accurate information of internal loading in various sitting conditions. But currently, owing to its dependence on vast amount of complicated calculation, most of the FE models reside on powerful computers in research facility far from being used in practical application. Therefore, methodology of simplification should be researched to produce **fast, simple, and handy models** which can be used in real life, such as in wheelchair fitting clinic. However, such simplification can only be properly done when the knowledge of buttock tissue/structure behavior in sitting is fully understood without biased knowledge which brought forth by premature simplification. Ideally, with successful simplification, a deliverable turn-key package (or packages) should be produced to professionals in the seating field for practical application for various purposes, e.g. wheelchair fitting, office chair design and evaluation, vehicle seat design, etc.

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