

Numerical Analysis of the Interactions between Human Body Soft Tissue and Body Supports

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Abstract: The number of patients in need of medical aid devices (anti-decubital systems, medical shoes, orthoses etc.) has risen drastically due to demographic changes and an unhealthy lifestyle. Such devices however, are not always helpful at affected soft tissue body sites. Sometimes they even aggravate the situation, leading to costly complications such as pressure sores, infection and amputation. Therefore, better assessment and understanding of device caused mechanical loading, involving critical stress and strain at the affected tissue region, is essential to more effectively design such devices. A method to approach such problems has been developed. Based on imaging techniques (MRI/CT) and 3D-CAD reconstruction techniques coupled with the inverse finite element method, data sets of the complete human body as well as of specific body regions have been generated. The particular biomechanical properties have been characterized in vivo. Additionally, body support system materials have been identified to numerically simulate realistic body/support interaction. Thus, with help of the Body Optimization & Simulation System (BOSS) and BOSS-models, mechanical stress and strain introduced by technical support systems to the skin and to deeper tissue levels (fat and muscle) in various body positions have been located. Varying stress and strain distribution, depending on the particular body support material and/or design has been quantified. The effects on the tissue due to different supports (cushions, mattresses, shoe soles etc.) have been compared. Critical body sites at the buttocks, when the patient is in a seated or recumbent position, as well as at the foot, under running/walking conditions, have been identified. The results provide a basis for biomedically oriented body support device optimization which aims to lower stress and strain at affected body sites.

Keywords: Biomechanics, Human Body Model, Human Soft Tissue Interaction, Discomfort, Pressure Sores, Tissue Stress and Strain Distribution.

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1. Introduction

Changes in lifestyle over the past decades have made comfort related questions in conjunction with seated or recumbent body posture and particular body supporting devices such as seats, mattresses, back rests, shoes etc. increasingly relevant. Body supports, designed for functionality in the past, are recently being considered from a different perspective. Besides functionality, design and material optimization are becoming more and more important to minimize discomfort.

Urgent action is required in the field of inpatient health care associated with sustained mechanical tissue loading. The bedridden elderly and those who suffer impaired sensation are especially affected. These patients cannot appropriately change their body position when exposed to prolonged tissue loading. As a result pressure sores may occur, promoting severe infection or even leading to death. Besides the severe trauma experience, pressure sores are immensely costly and impose a significant burden on the health care system. Annual costs are estimated at \$4 billion for the UK “(Bennett, 2004)” alone.

Although pressure sore development is multifactorial, it is commonly agreed that sustained mechanical tissue loading plays the most significant role during sore formation. The perception of discomfort is a preliminary stage of a process leading to skin irritation and, if not resolved, may lead to pressure sores. Hence, skin level interface stress is important. Internal tissue stress and strain, which play a key role in understanding the effects of a particular supporting device on the body, are even more important. Lowering stress and strain at critical body sites represents a first step towards minimizing discomfort and preventing sore development.

Analysing mechanical stress and strain distribution inside human soft tissue to optimize relief and increase durability and reliability of body support devices is gaining importance. Concomitantly, the potential to derive and model detailed human anatomy with suitable in vivo tissue properties for use in computational finite element (FE) body/support interaction simulation is evolving. In the FE simulation process, adequate material parameters for both soft tissue and support material are indispensable to properly describe mechanical behaviour.

The human buttock region has been subject to several computational studies investigating the stress-strain state of gluteal soft tissue when in the supine or sitting position. The complex tissue compound was reduced to skin, fat, muscle and bone which can be properly differentiated using magnetic resonance imaging (MRI). The material descriptions of human gluteal soft tissue within these studies were based on material parameters which were derived in different manners. Specifically, buttock indentation tests have been performed “(Todd, 1994)” and the soft tissue compound was assumed to be linearly elastic. Oomens et al. “(Oomens, 2001)” used rat data to represent human muscle tissue and skin parameters derived from pig experiments. Fat parameters were assumed. In an additional study by Oomens et al. “(Oomens, 2003)”, again, animal data were employed as a basis for the investigation. Using a neo-Hookean material model, Sun et al. “(Sun, 2005-A; Sun 2005-B)” made assumptions about the material properties of skin, fat and muscle assuming weak tissue compressibility. Literature based parameters were used in recent studies by Lim et al. “(Lim, 2006)” and Makhsous et al. “(Makhsous, 2007)” using the Mooney-Rivlin material model to describe human soft tissue behaviour. Linder- Ganz and Gefen “(Linder-Ganz, 2007)” used a neo-Hookean material model employing material data of porcine gluteal muscles as

well as skin/fat data from indentation testing. The 3-dimensional anatomical structures of the human buttock used within these investigations were mostly based on reconstructed MRI-data, e.g. “(Sun, 2005; Lim, 2006; Makhssous, 2007)” which provided realistic anatomical information. More abstract models of the buttock were used in “(Todd, 1994; Oomens, 2003)” aiming to account for the stress-strain state within the tissue adjacent to prominent bone structures. Tissue layer thickness information herein as well as abstracted anatomical structures were derived from MR-images. Besides the employed geometric modelling approaches, the studies above have mainly been based on tissue parameters derived from ex vivo animal experiments. This is due to a paucity of information on human material parameters, characterizing in vivo human gluteal fat and muscle tissue properties.

The material parameters used in the present investigation for simulating human body/ support-interaction in supine or sitting body position represent in vivo human gluteal skin/fat and in vivo human passive muscle tissue behaviour. The rate-independent long-term elastic material properties for gluteal skin/fat and passive muscle tissue have been characterized independently. A combined iterative numerical-experimental approach based on in-vivo non-invasive testing as described in “(Then, 2007)” was employed. The constitutive equation for non-linear, hyperelastic, isotropic materials proposed by Ogden “(Ogden, 1972-B)” was used to represent tissue behaviour. To independently account for time-dependence of both tissue types, viscoelastic material parameters for gluteal skin/fat and muscle tissue have been derived. Again, a combined iterative numerical-experimental procedure based on in-vivo non-invasive creep testing at the buttocks was employed. Prony series parameter sets for both tissue components were established separately. Similarly to methods employed for human soft tissue, body support material characterization, mainly soft polymeric foams, was performed. The modified Ogden-Hill model “(Ogden, 1972-A; Ogden, 1972-B; Hill, 1978)”, following the approach described in “(Schrodt, 2005)” to extract the long-term material response was applied. Assuming that the material behaviour of both interacting partners, tissue and support, is dominated by a creep process, creep tests were performed.

Body/mattress-interaction was simulated using a body model in the supine position. For body/seat cushion-interaction a model in a seated position was used. Body anatomy in both cases was derived from MR-images from subjects in the incumbent and seated positions, respectively. Depending on the particular support material and/or design, tissue stress and strain distribution was evaluated and supports were relatively compared. Critical body sites in both body positions were identified and support design optimization may be used to lower stress and strain in these regions. In general, it has been shown that maximum strain in the gluteus region occurs in tissue near bony prominences (ischial tuberosity; posterior superior iliac spine; sacral bone edges), not at the skin. In the majority of cases, interface stress at the skin level is up to 50% lower than internal stress near the bone surface. These findings emphasize the need to account for internal tissue stress-strain distribution. Hence, common interface pressure evaluation tools such as pressure mapping systems must be questioned.

2. Materials and Methods

The experimental protocols applied to volunteers participating in tissue testing were approved by the ethical committee of the Hospital of the Johann Wolfgang Goethe University of Frankfurt. Informed, written consent was obtained.

In the following, material characterization of both interacting materials, i.e. human gluteal soft tissue and foam support material is briefly described. Both materials showed viscoelastic material behaviour during testing. To describe the long-term non-linear elastic material behaviour, constitutive equations for hyperelasticity are permissible. As employed in this investigation, the constitutive equations in terms of the principal stretches used for human soft tissue “Equation 1,” and soft polymeric foams, “Equation 2,” read

(1):

$$\mathbf{S}_0 = 2[J(\mathbf{X}, t)]^{-1} \sum_{i=1}^3 \sum_{k=1}^N \left\langle \frac{\mu_k}{\alpha_k} [J(\mathbf{X}, t)]^{\frac{\alpha_k}{3}} \left\{ [\lambda_i(\mathbf{X}, t)]^{k_k} - \frac{1}{3} \sum_{j=1}^3 [\lambda_j(\mathbf{X}, t)]^{k_k} \right\} + \frac{k}{D_k} J(\mathbf{X}, t) [J(\mathbf{X}, t) - 1]^{k-1} \right\rangle \mathbf{n}_i \mathbf{n}_i$$

(2):

$$\mathbf{S}_0 = 2[J(\mathbf{X}, t)]^{-1} \sum_{i=1}^3 \sum_{k=1}^N \frac{\mu_k}{\alpha_k} \left\langle [\lambda_i(\mathbf{X}, t)]^{k_k} + \frac{1}{\alpha_k} J(\mathbf{X}, t) \frac{\partial f[J(\mathbf{X}, t)]}{\partial J(\mathbf{X}, t)} \right\rangle \mathbf{n}_i \mathbf{n}_i$$

with the eigenvalues λ_i of the right stretch tensor $\mathbf{U}$, the eigenvectors $\mathbf{n}_i$ of the left stretch tensor $\mathbf{V}$, the material coefficients μ , k , α_k and β_k ($k= 1, 2, \dots, N$) and $J = \det \mathbf{F}$.

Based on the constitutive equation for linear viscoelasticity at finite strains, “Equation 3,” the time-dependant part of the particular material was determined performing specific creep tests. The Prony series relaxation parameters were defined by an iterative optimization procedure using ABAQUS[®] as the exact analysis solver to provide the particular model functions. These were fitted to the test data.

(3):

$$\mathbf{S}(\mathbf{X}, t) = \mathbf{S}_0(\mathbf{X}, t) + \int_{t'=0}^t \frac{\dot{G}(t')}{G_0} \frac{J(\mathbf{X}, t-t')}{J(\mathbf{X}, t)} \mathbf{S}_0(\mathbf{X}, t-t') dt' \quad \text{with} \quad \frac{\dot{G}(t)}{G_0} = - \sum_{i=1}^M \frac{g_i}{\tau_i} e^{-\frac{t}{\tau_i}}$$

where G_0 is the instantaneous shear modulus, $G(t)$ is the time-dependant shear relaxation modulus defined by a Prony series expansion, g_i , τ_i are material constants and $\mathbf{S}_0$ given in “Equation 1,” and “Equation 2,” respectively.

Human Gluteal Soft Tissue Characterization

Material characterization of human gluteal skin/fat and passive muscle tissue was carried out and separate parameters sets describing in vivo fat and muscle tissue properties were established “(Then, 2007)”. Indentation loading at the buttocks was performed and indentation force and indentation displacement were recorded. To obtain information about the displacement field inside the tissue, tissue deformation was performed stepwise and tissue displacement was recorded through MR-imaging. The stepwise procedure assured separation of the elastic from the inelastic material properties. The basic assumption of this approach was that the current tissue displacements along the indentation axis under loading were digitally measured from the particular MR-image, for example “Figure 1b,” and then were referenced to the undeformed tissue configuration, “Figure 1a”. Thus, individual force-displacement relations describing gluteal fat and muscle behaviour were established. The key to describing this relationship is ensuring defined boundary conditions within the deformation process. This was accomplished with an indentation axis position, (position [1] in “Figure 1a”) meeting the pelvic bone surface, centred and orthogonal.

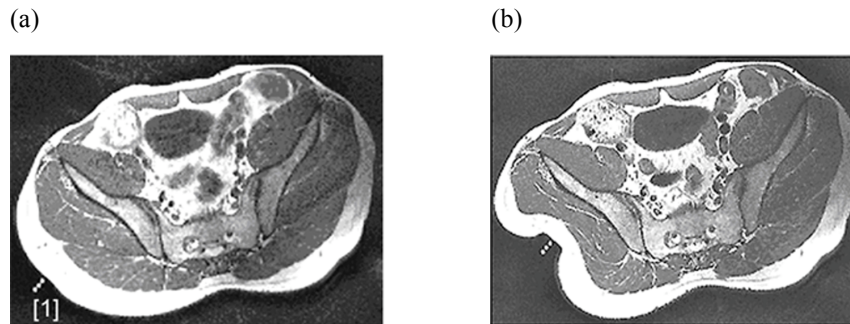


Figure 1. (a) Transversal MR-image of the gluteus in the initial undeformed state, (b) At maximum indenter displacement.

In the tissue parameter optimization process the deterministic simplex strategy optimization algorithm was employed. The material coefficients given in the Ogden form for slightly compressible, isotropic, hyperelastic materials were independently parameterized and optimized for both tissue components. The established force-displacement relations for fat and muscle tissue thereby reflect the test data and served as target functions. Material stability in the sense of the Drucker stability condition was assured for the derived soft tissue parameters for the strain state resulting from the particular experimental procedure. In this context, the Drucker stability criterion was assigned as an additional constraint in the optimization process. The aim thereby was to maximize the particular strain levels reflecting stable material behaviour for the general deformation modes “(ABAQUS, 2008)”.

Subject studies revealed that during indentation, progression of the force-displacement curves of the fat and muscle tissue have an inherent characteristic. Whereas both tissue types show a nearly linear initial force progression, fat tissue suddenly blocks with the force rapidly rising at

small strain increments. Muscle tissue, however, exhibits moderate force increase at increasing strains.

Determination of the time-dependent linear viscoelastic material parameters of gluteal fat and muscle tissue was done by performing creep testing with MR-imaging. A 50% higher creep deformation in the transversally loaded passive gluteal muscle tissue, compared to gluteal fat tissue was revealed. Following a similar approach, as previously described, indentation at the buttocks was performed at constant force using the cine-MR-imaging method. Displacement field information as a function of time for both tissue types was thereby obtained. Performing MR-image reconstruction, individual displacement-time relations for each tissue material were extracted. An FE-model of the initial undeformed configuration was constructed, based again, on the MR-imaging information. Since cine-MRI provided image data in the indenter axis plane only, three distinct material points in the fat/muscle interface were selected. For each point an individual displacement-time relation was established. Together with the constant indentation force constraint, the relation served as boundary conditions in an iterative optimization process. Viscoelastic material constants of fat and muscle tissue were parameterized and optimized simultaneously.

Human Gluteal Soft Tissue Material Parameter Validation

Material parameter were validated with a defined loading scenario employing MR-imaging as described in “(Then, 2009)”. Again, to gain displacement field information, MR-images of the loaded configurations were obtained and visually compared with simulation results. Deformation of gluteal skin/fat and passive muscle tissue and support material under interacting loading, derived by numerical simulation, was in good agreement with the corresponding MR-image results. Accordance was found for the deformed skin surface as well as the internal fat-muscle tissue boundary surfaces by superimposing experimental and numerical outputs, “Figures 2a,” and “Figure 2b,” respectively. In addition, a new concept of FE-model validation regarding non-MR sensitive materials such as polymeric foam was introduced, including peripheral surface visualization of the specimen.

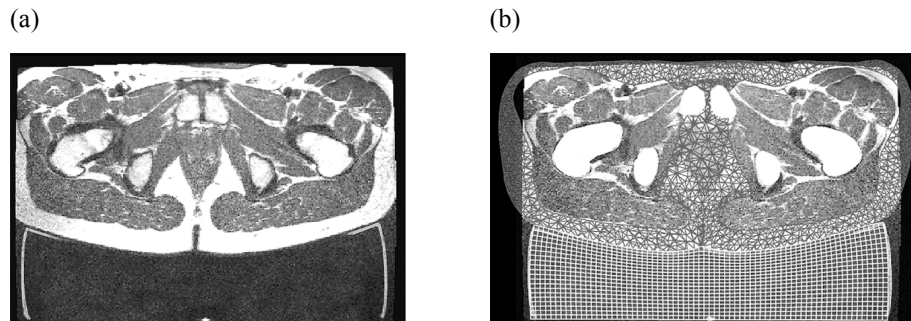


Figure 2. (a) Transversal MR-image of the loaded buttocks with foam test specimen (side face contour) at maximum loading, (b) Superposition of MR-image and FE-mesh.

Human Heel Tissue Characterization

Following the same approach described earlier, heel tissue was characterized via long-term and time-dependent material characteristics. A specific, MRI compatible loading device was developed to simultaneously load the heel pad and record force-displacement and displacement-time relations.

Polymeric Soft Foam Material Characterization

Mechanical characterization of open-cell polymeric soft foam material followed a stepwise loading procedure, employing a uniaxial rigid plate and hemispherical rigid indenter testing. This characterization was initially proposed by James & Green "(James, 1974)," successfully applied by Hartmann et al. "(Hartmann, 2003)" and Lion "(Lion, 1996)" for rubber-like materials and Schrodtt et al. "(Schrodtt, 2005)" for polymeric foams. The experimental long-term force-displacement material responses obtained by separating the elastic from the inelastic material properties served as object functions in an iterative numerical optimization process. Foam material parameters could thus be derived. In order to numerically reproduce both indentation tests adequately, in the majority of cases it was essential to perform a multi-objective parameter optimization with all objectives being mutually in conflict. Uniaxial and multiaxial compression test data were considered to finally derive one agreeing parameter set, satisfying both test conditions.

In the parameter optimization process a material form for highly compressible isotropic hyperelastic materials was used to model long-term soft foam properties. This form was first introduced by Hill "(Hill, 1978)" and applied to highly compressible soft foams in an experimental work by Storåkers "(Storåkers, 1986)". Again, the Drucker stability condition was assured for the experimentally experienced strain range.

Determination of the time-dependent linear viscoelastic material parameters of polymeric soft foams was done by performing uniaxial and hemispherical indentation creep testing. Again, the Prony series parameters were parameterized and optimized by iteratively simulating the test scenarios. For verification of the long-term and the time-dependant material parameters, an authentic buttock model made of synthetic resin was built and creep testing was performed. The scenario was FE modelled and simulated and the displacement-time data compared to the test output. The investigated foam materials were sufficiently characterized using a linear viscoelastic approach.

Body Part Modelling and Body/Support Interaction

With help of the Body Optimization & Simulation System (BOSS)-procedure, appropriate material parameters for human tissue and support material were derived. Based on MR-imaging data, anatomically adequate FE- body models were constructed. These included buttocks when in the supine and seated positions, as well as the foot region. Fat, muscle and bone tissue were differentiated, "Figure 3a," and "Figure 4a". Body kinematics were accounted for by including leg

and spine structure interconnected with joints. Vertebrae were assigned moment-angle characteristics to allow relative motion.

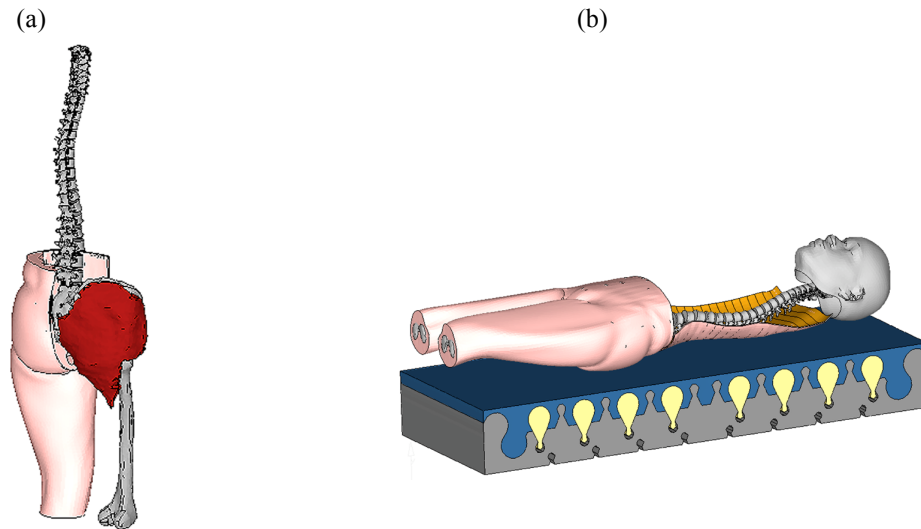


Figure 3. Recumbent BOSS-body model: (a) including bone structure, gluteal muscles and fat tissue (cross-section), (b) Body/Mattress support interaction.

Tissue parameters were assigned to gluteal fat and muscle tissue. Bone parts were assumed to be rigid and lumped masses were added to the particular body parts to meet average weights. Male and female anatomy was reproduced, including appropriate tissue parameters for both sexes. Static body weight loading on various supports, “Figure 3b,” as well as static and dynamic car seat vibration loading, “Figure 4b,” was simulated. Tissue stress and strain were then evaluated.

(a) (b)

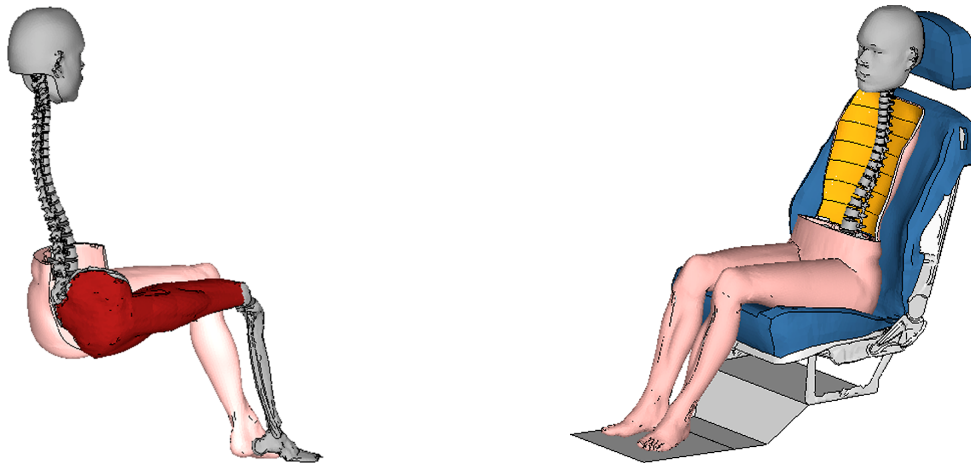


Figure 4. Seated BOSS-body model: (a) including bone structure, gluteal and thigh muscles and fat tissue (cross-section), (b) Body/Automotive car seat interaction.

The stress and strain situation inside human heel tissue was analyzed during gait loading, subject to various shoe sole materials. To capture walking kinematics, experimental gait analysis, “Figure 6a/b,” was performed where the spatial motion at specific marker points was captured and mapped on an FE-human foot model, “Figure 5” to simulate the walking/running process.

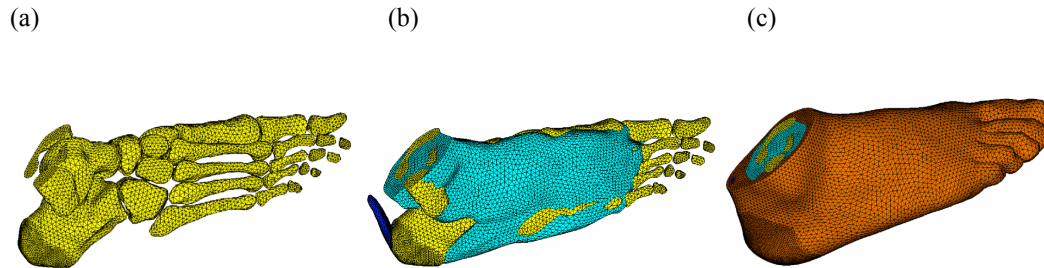


Figure 5. (a) Bone structure, (b) Muscles and ligaments, (c) Fat tissue.



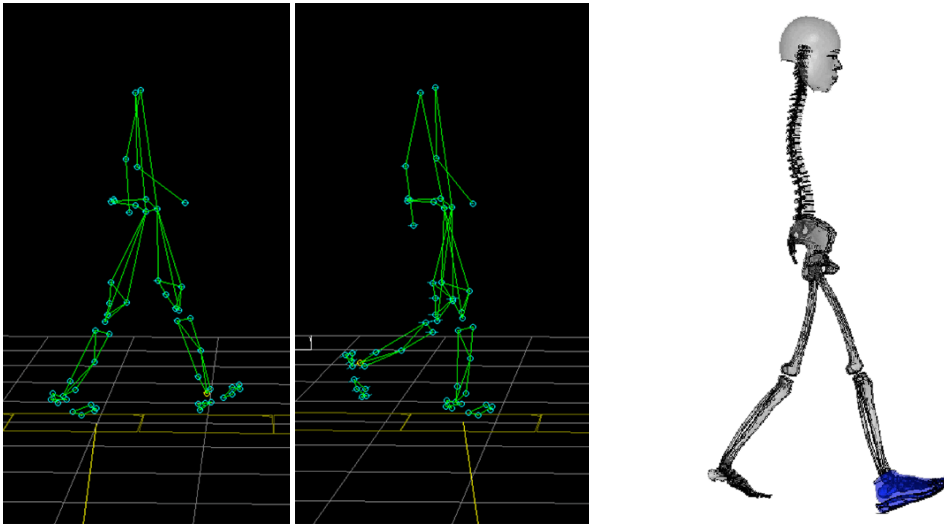


Figure 6. Reconstructed marker signals of gait analysis during (a) Walking motion, (b) Running motion, and (c) FE gait simulation.

The current foot model consists of the basic bone structure, “Figure 5a,” partially interconnected with 13 joints. Bone groups that exhibit relatively small relative motion have been considered as rigidly attached. Additionally, basic bone motion is restricted by 18 ligaments.

3. Results

The body models were simulated in supine and seated positions on various supports. Interface well as internal tissue stress and strain were then evaluated. Increase of internal tissue stress and strain can be explained taking the bone structure, tissue material characteristics and external loading situation into account.

Static Body/Mattress Support Interaction

A comparison of the effects of different support designs/materials on the interface and internal tissue stress distribution was performed “(Figure 7, top)”. Distinctive anatomic points, the ischial tuberosity, the posterior superior iliac spine, the lateral sacral bone edge and the lateral tail bone, were particularly prone to stress accumulation. Interaction of compliant support material and soft tissue makes internal tissue stresses and strains increase near prominent, irregular bone surfaces during compression. Here, soft tissue is restricted at the relatively small, prominent bone surface and adjacent tissue material is displaced ‘around’ the particular bone prominence, as marked between the ischium [1] and the femur [2] bone in “(Figure 7, bottom)”. This effect causes tissue strain to increase below the prominent bone region and stress to increase at the lateral region of the prominence.

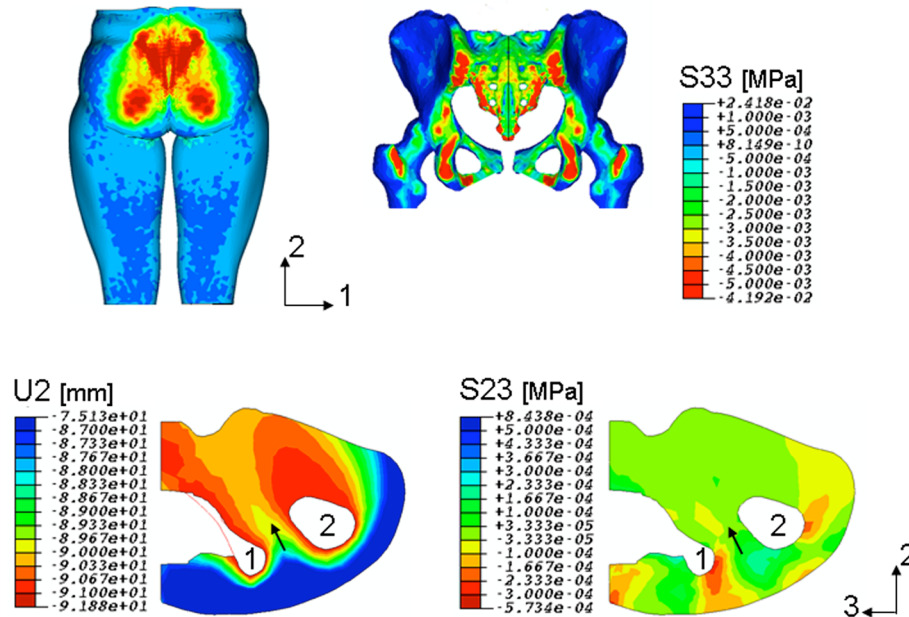


Figure 7. Simulations of body/support interaction. Top: compressive stress at the skin interface and at the bone. Bottom left: tissue displacement due to body weight loading and resulting shear stress maxima at the ischium, bottom right.

Dynamic Body/Seat Support Interaction

Simulations of the seated body model, “Figure 8a,” on various seat supports show a maximum stress increase from the skin towards the ischial tuberosity. Again, as when in a recumbent position, a strain maximum is found inside the muscle tissue below the prominence of the ischium, cf. enlargement in “(Figure 8a)” showing shear stress component S23. At the skin level, the region below the ischium shows maximum compressive stress, “Figure 8b”. In contrast to the recumbent position, most of the upper body weight is distributed via the relatively small buttock/seat interface. As a result, compressive stress at the ischium is approximately double the stress magnitude found when lying down.

Stress and strain distribution inside the tissue for different seat designs and materials have been compared. Based on these results, basic principles of seat cushion design have been established.

(a)

(b)

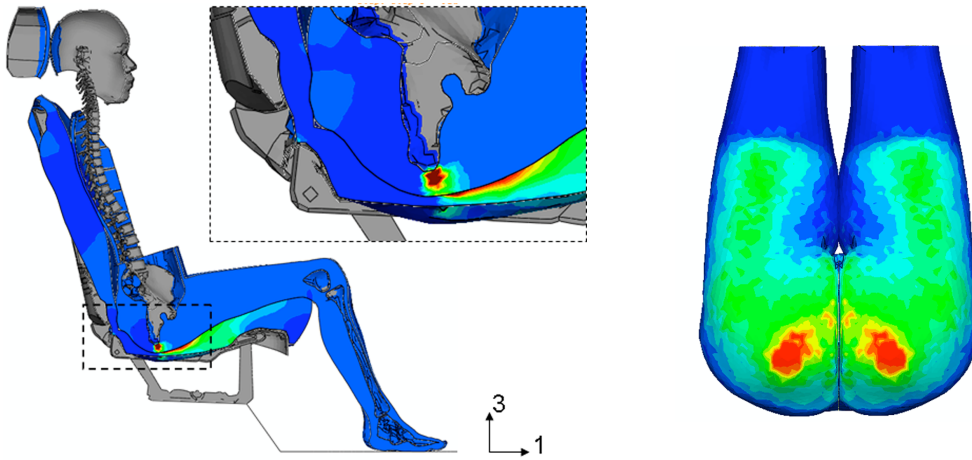


Figure 8. Simulation of body/support interaction when seated in a car seat, (a) half-section cut side view at the ischium , (b) buttocks bottom view.

Impact Heel/Shoe Interaction

Human gait analysis reveals stress and strain distribution inside the heel tissue during heel impact, “Figure 9”. Depending on the particular shoe sole material and sole design, maximum values vary. Relative comparisons of different design shapes were simulated, and the effects on the internal stress/strain state were evaluated. Comparison between bare-footed walking/running and walking/running with various shoe sole designs/materials has been simulated. The effect of the particular shoe sole could thus be clearly differentiated. Gait analysis is extended to include upper and lower leg interconnected with joints as well as soft tissue to analyze force/moment transmission in addition to stress and strain distribution.

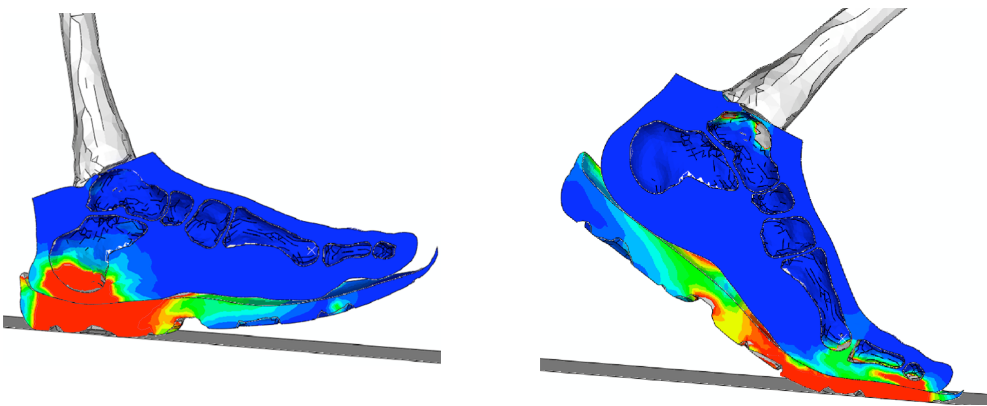


Figure 9. Human gait analysis providing insight into the process of heel impact.

4. Discussion

FE-simulation of human tissue and support device interaction in seated and recumbent body positions as well as under walking/running conditions has been performed. Thereby insight has been provided into internal mechanical tissue processes during particular loading situations. Comparing effects of different support designs/materials on tissue have helped to acquire a more profound understanding of the internal tissue stress/strain distribution. Thus, it becomes increasingly possible to optimize support devices in a particular loading condition. Minimizing internal tissue stress and strain results in a reduction of discomfort. Pressure sores can be prevented and the time between external repositioning of immobilized patients can be extended. FE-simulation is superior to pressure mapping systems, which are clearly inadequate in evaluating the stress/strain distribution inside human tissue. As has been proposed in previous investigations “(Todd, 1994; Oomens, 2003; Then, 2008; Silber 2007),” the finding that interface stress generally exceeds internal stress questions the usefulness of interface body pressure measurement systems. These have commonly been used for pressure measurement of seated or recumbent subjects to evaluate the anti-decubitus effectiveness of body supports. Such systems permit rudimentary estimation of mechanical pressure to the skin surface but do not capture interface shear stress or internal tissue stress, let alone internal tissue strain.

Tissue movement during loading is the factor causing strain to increase inside the tissue when in the supine or seated position. Therefore, support design optimization should focus on restricting intense tissue movement relative to the bone. The bedding and seat industry has developed mattress and cushion designs with a planar contact surface to the skin. This does not take the body shape e.g. at the buttocks nor the shape of the internal human bone structure into account. Manufacturers claim to have developed effective anti-pressure sore or comfortable products with various fancy support design shapes and materials. Currently, materials with viscoelastic supporting properties are top-selling. However, no positive evidence concerning internal tissue conditions from a mechanical point of view exists. Finite element simulations of body/support interactions employing adequate description of interacting human soft tissue and support material will help understand the influence of such support materials. This, in turn, can contribute to a much more effective optimization process.

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