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Basic research

An air-cell-based cushion for pressure ulcer protection remarkably reduces tissue stresses in the seated buttocks with respect to foams: Finite element studies

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Abstract A sitting-acquired pressure ulcer (PU) is a common injury in wheelchair-bound patients. Preventative measures for the post spinal cord injury (SCI) population include prescription of a supportive thick cushion on the wheelchair, in order to better distribute loads between the buttocks and support surface (which are quantifiable using interface pressure measurements), and potentially, to minimize internal soft tissue loads (which are typically unknown). Information about the biomechanical efficacy of commercially-available structured cushion designs such as air-cell-based (ACB) cushions, gel, and honeycomb-like cushions is sparse. Considering the importance of such evaluations to patient safety and quality of life, we studied the biomechanical performances of an ACB cushion in comparison to standard, flat foam cushions with different stiffness properties. Using a set of finite element (FE) model variants, we determined the mechanical stresses in muscle, fat, and skin tissues under the ischial tuberosities during sitting. Tissue stress analyses were conducted in a reference SCI anatomy, incorporating pathoanatomical and pathophysiological changes associated with chronic SCI, including bone shape adaptation, muscle atrophy, and spasms. We found up to 57% greater immersion and 4 orders-of-magnitude lower muscle, fat, and skin tissue stresses for the ACB cushion. We also found the ACB cushion provides better protection against the aforementioned bone shape adaptation, muscle atrophy, and spasms. Hence,

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theoretically, the use of a suitable ACB cushion should provide longer safe sitting times for SCI patients with respect to standard foam cushions.

Introduction

A sitting-acquired pressure ulcer (PU) is a common complication in wheelchair-bound patients after a spinal cord injury (SCI) [10,13]. Sitting-acquired, deep tissue injury (DTI) is a severe form of PU which onsets under intact skin, usually near the interface between the ischial tuberosity (IT) and overlying soft tissues, and is also characteristic to SCI patients [2,16]. In community settings in the U.S. and Europe, the incidence of PUs in SCI patients is estimated between one-quarter and one-third of the population [9,23]. Annual PUs treatment costs to the U.S. healthcare system are estimated at 11 billion dollars, and the total cost to manage a single full-thickness PU can be as high as $70,000 [5].

Following a SCI, pathoanatomical and pathophysiological changes occur in the buttocks, as tissues adapt to the chronic sitting and inactivity of muscular innervations. These changes include bone shape adaptation (loss of cortical bone and flattening of the tips of the ITs) as well as muscular atrophy and sometimes muscle spasms [3,4,12,24]. These phenomena are likely to affect the risk of PUs in SCI patients. As the alterations of the weight-bearing structures occur, the internal loading states in the tissues are affected.

Due to the clinical challenge and costs in treating PUs, tremendous efforts are being made towards efficient risk assessment and prevention strategies [7], which greatly depend on adequately understanding the etiology. The most important guidelines for prevention of sitting-acquired PUs is to use a soft, but thick-enough cushion on the wheelchair to better distribute the buttocks-support loads while preventing undesirable bottoming-out [7]. Contact pressure measurements are unable to demonstrate internal soft tissue loads in the buttocks. Hence, this method is insufficient for fully assessing the biomechanical performances of cushions. However, the use of finite element (FE) modeling provides a complementary research methodology, which can evaluate internal soft tissue loads, and isolate the contributions of parameters associated with specific cushions or patients, to the risk of developing PUs [1]. Several studies used FE modeling to investigate the biomechanical performances of basic foam cushions (single-type, flat and uniform material) [17,18,21,22,27,30,31] however, more complex cushion structures were not evaluated through FE. The variety of commercially-available wheelchair cushions, such as air-cell-based (ACB) cushions, gel, and honeycomb-like cushions calls for expanding the use of FE modeling to these complex cushion structures.

Here we used FE modeling, to computationally evaluate, for the first time, how a more sophisticated structured cushion affects internal tissue loads. We focused on the ACB cushion, and used foam cushions for comparison. We studied muscle, fat, and skin stresses during sitting, in a reference SCI anatomy versus model variants representing the above SCI-related adaptations to the buttocks' structure. The work demonstrated how FE modeling can serve the cushion industry, by adding the specific knowledge relevant to minimizing DTIs.

Methods

The FE method is a computational technique for finding the internal mechanical loads, (deformations, strains and stresses) in structures having complex shapes and multiple materials. In practice, the complex geometry of the structure is divided into numerous small elements — each with a simple geometry (such as pyramids), and the differential governing equations that describe the mechanical interactions are solved numerically for every element with respect to its neighboring elements, in order to ultimately construct the solution to the entire structure.

Geometry

In order to examine the effects of support stiffness and altered anatomical structure post a SCI, in particular IT flattening, gluteus muscle atrophy and spasms, a set of 15 model variants of the left buttock were developed (Table 1). Each model variant was based on a coronal cross-section, and included the IT bone, the gluteus maximus skeletal muscle, the colon smooth muscle, fat tissue, skin, and either a flat foam cushion or an ACB cushion (Fig. 1). The model variants differed in foam cushion stiffness, gluteus muscle mechanical properties (representing normal versus spastic muscles), and pathoanatomical adaptations of the muscle and bone, as detailed in Table 1.