The Critical Characteristics of a Good Wheelchair Cushion

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Introduction

Pressure ulcers (PUs) are known to develop when soft weight-bearing tissues are subjected to sustained increased deformations, usually between a bony prominence and an external support surface [1–3]. PUs are commonly staged with respect to the depth on the ulcer and the tissues it involves, with the most severe ulcers, which involve muscle and bone tissues, termed deep tissue injuries (DTI). A number of contributing or confounding factors, such as impaired mobility and sensory capacities, alterations in skin status and moisture, poor nutrition and impaired perfusion, are also associated with PU development [2, 4]. Hence, populations at risk are the elderly and frail, patients post spinal cord injury (SCI), brain trauma or stroke, patients with impaired mobility or sensory capacities, or even patients who undergo prolonged surgery, as these individuals are more likely to spend prolonged periods in a static position, and are also less like to readily detect the risk [2]. The prevalence of PUs in the acute, critical and pediatric care settings can be as high as 46%, 45.5% and 72.5%, respectively, while the incidence of PUs in the aged care settings can be as high as 59% [5]. The most prevalent locations for PU development are the sacrum (28.3%), buttocks (17.2%) and heels (23.6%) [6], which are associated with both prolonged sitting and supine lying. The average monthly cost per such case, e.g. to the Canadian healthcare system, was reported to be $4745 [7]. The total cost to manage a single full-thickness PU in the United States can be as high as $70,000, and PU annual treatment costs to the US healthcare system are estimated to be 11 billion dollars [8]. Many studies have documented the increased morbidity and

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mortality associated with PUs in both community and hospital settings, as well as their significant contribution to healthcare costs and lengths of hospital stay [9].

There is evidence in the literature that PUs and PU interventions have a significant impact on health-related quality of life and inflict substantial burden on patients [10]. Reported that the majority of patients indicated that suffering PUs affected their lives emotionally, mentally, physically and socially [11]. While the most common complaint is the pain, patients also express their discomfort with the appearance, smell and fluid leakage from the ulcer. PU patients are often dependent on others to manage and care for their ulcer and describe their discomfort with the dressing materials and pressure-relieving equipment [11]. Since treatment of PUs is medically challenging, costly, and very unpleasant for sufferers, efforts are put towards risk assessment and prevention strategies [2], which in turn, depends on thorough understanding of the aetiology.

Pressure Ulcers in the Spinal Cord Injury Population

Sitting-acquired PUs are common in individuals who chronically depend on a wheelchair for mobility, such as those with a SCI. These patients spend up to 18 h a day in a wheelchair and often suffer from impaired sensation in their buttocks, preventing them from detecting the risk in a timely manner. In fact, in Europe as well as in the U.S., at least one in every four persons with SCI is affected by PUs, with the most common site being under or around the ischial tuberosities (ITs) [12, 13].

In the weeks and months following the acute injury, disuse-related pathoanatomical and pathophysiological changes occur in the buttocks, as tissues adapt to the chronic sitting and inactivity of muscular innervations. Perhaps the most documented is the disuse-induced muscular atrophy (MA), which includes massive loss of muscle volume [14], thinning of muscle fibers, and increased numbers of fast-twitch over slow-twitch fibers. The MA onsets as early as 4–6 weeks after the acute injury, and progresses at a noticeable rate for at least several months, after which the rate of muscle wasting tends to slow down, but the absolute tissue loss persists [15–17]. High levels of intramuscular fat infiltration (FI) are also prevalent after a SCI. Intramuscular adipose depots can be found in able-bodied individuals as well, however, while the normal intramuscular FI level is 1–2% of the total body fat [18], FI levels post SCI can be up to 4-times greater than that. The chronic sitting and disuse also affect the weight-bearing bony structures. Specifically, substantial bone loss has been described in a number of cross-sectional studies, with the time course of bone loss depending on the bone compartment. Rapid loss of trabecular bone may level off 1–3 years post injury, while slower cortical bone loss appear to decrease progressively beyond 10 years post injury [19]. Furthermore, bone shape adaptation (BA), namely flattening of the tips of the ITs in response to the sustained sitting loads has been reported in the literature [20–22]. 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.

Furthermore, initial weight-loss followed by considerable weight-gain are very common in the few months and years following a SCI. Specifically, there is tendency for an initial weight-loss of 5.3–9.1 kg at the short term (within weeks) after the acute injury, which is followed by a major weight-gain of 1.3–1.8 kg *per week* during the rehabilitation phase [23, 24]. The decrease and increase in the body-weight are due to a hypercatabolic 'shock' response and a lower-level of physical activity, respectively [25].

In addition, one may consider a patient who already had PUs in the past, which healed but left scars in the skin, fat and/or muscle tissues, and hence the mechanical properties of these soft tissues of the buttocks are locally abnormal and inhomogeneous, which in turn affects internal tissue loads in the scars and also in adjacent, non-scarred tissues.

Computational Modeling for Studying the Efficacy of Wheelchair Cushions

The finite element (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.

In order to examine the effects of (age-related) skin stiffness, soft tissue scarring, bone shape-adaptation, MA, FI and changes in BMI post a SCI, on the mechanical loads developing in the soft tissues of the buttocks of a SCI patient, 54 model variants of the seated buttocks were developed [26–29] (Fig. 2.1). Each model variant included the ischial tuberosities (ITs), the gluteus maximus skeletal muscles and the colon smooth muscle, subcutaneous fat tissue, skin and either a flat foam cushion, a contoured foam cushion (CFC) or an air-cell-based (ACB) cushion for support. The model variants differed in support structure and stiffness properties, IT anatomy, fat and muscle volume and structure and soft tissue global stiffness or local scarring. Each of the model variants was based on a single, coronal MRI slice through the buttocks, acquired from a male subject 1 year after a SCI (age 21 years, weight 90 kg) who was scanned in an open MRI, sitting on a rubber tire (non-weight-bearing) and then fully weight-bearing on a semi-rigid flat support in our previous work [22] (Fig. 2.1). To generate the reference anatomical model, the non-weight-bearing MRI slice was loaded to the ScanIP® module of Simpleware® [30], where it was automatically segmented to the different tissue components listed above and then uniformly extruded to a 4-mm depth, representing the MRI resolution in the *Z*-axis. Mechanical properties of all tissues were adopted from the literature [26–29].

Next, we artificially introduced pathoanatomical variations and different supports to form the different model variants (Fig. 2.1). First, we investigated how thin, flat or hypertrophic scars in the skin affect the developing soft tissues shear loads

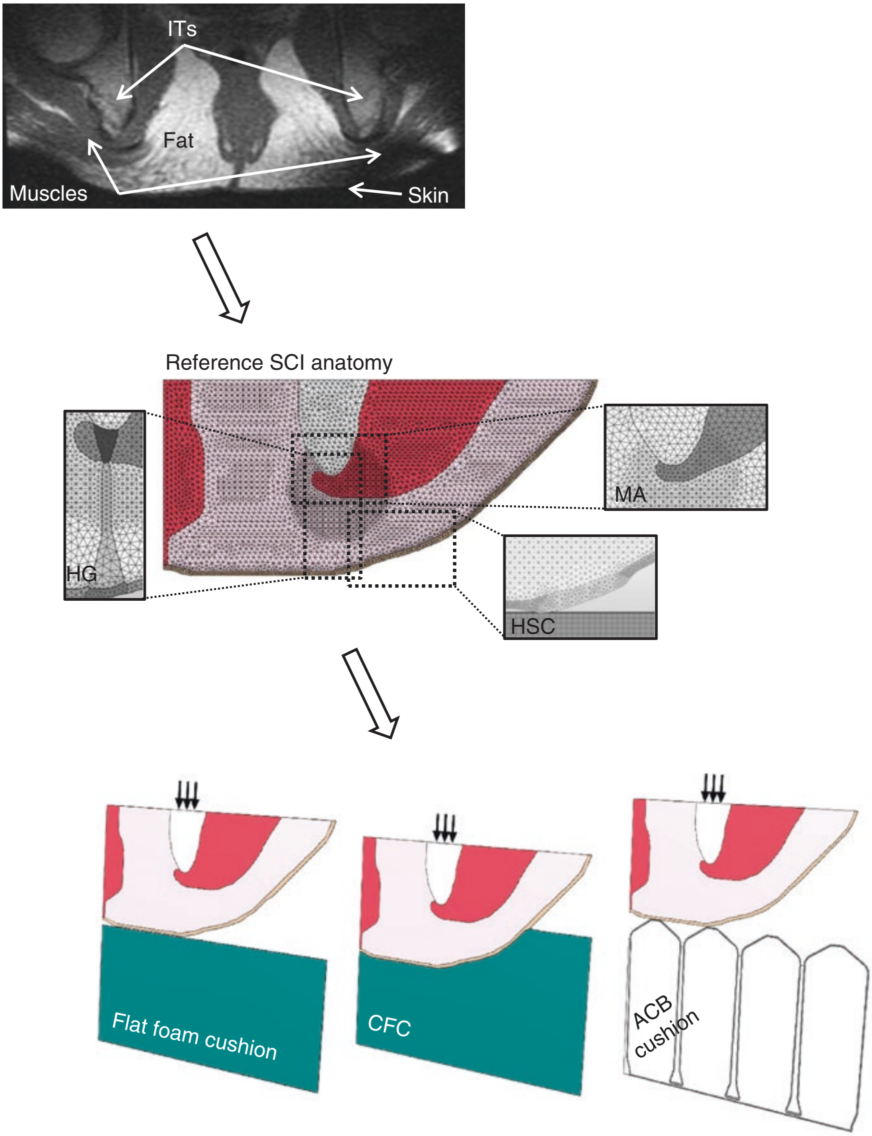

Fig. 2.1 Computational (finite element) modeling of the buttocks of an individual with a spinal cord injury (**SCI**): (**a**) Anatomical components of the model variants, as seen in the MRI slice. (**b**) Three-dimensional mesh of the reference anatomy and three of the considered pathoanatomical variations, for example: (1) ***HG*** hourglass-shaped scar in the muscle, fat and skin tissues, (2) ***MA*** muscular atrophy, (3) ***HSC*** hypertrophic skin scar. (**c**) General loading configurations of the buttocks when seated on a flat foam cushion, a contoured foam cushion (**CFC**) or an air-cell-based (**ACB**) cushion

during sitting-down on flat foam cushions with different stiffness properties, in 'young' or 'aged' skin conditions [26]. Adding these skin scars into the model allowed investigations of how shear loading may develop when the skin is supposedly more fragile or already locally damaged. Next, we examined how BA, MA, muscle spasms or a combination of the above, affect peak soft tissue stresses when seated on two flat foam cushions or an ACB cushion [27]. The geometry of the ACB cushion was based on a 4-mm slice through the pre-inflated air cells and its mechanical properties were evaluated using a simple buckling experiment detailed in [27]. Then, we looked into how muscle, fat and skin tissue scarring in a patient with a history of PUs in their buttocks, affect the resulting mechanical stresses in these soft tissues during sitting on the aforementioned ACB cushion [28]. We introduced ten scars of different shapes and dimensions to the model variants, corresponding to the modeling work of Sopher et al. [31], to describe cases of patients who already suffered PUs in their buttocks, which healed but left scars in their soft tissues. Finally, we explored how changes in BMI, intramuscular FI, MA and combinations of these conditions, affect the internal soft tissue loads in the buttocks of SCI patients, while sitting on a CFC, which has been fitted close to the time of the injury but has not been replaced in subsequent years [29].

Loading configurations were chosen to simulate descent of the ITs downwards, as these bones transfer a portion of the bodyweight to the overlying soft tissues during sitting. When tested on flat foam cushions or the CFC, uniform pressure was applied downwards on the top of the ITs with its magnitude calibrated by fitting the resulting vertical displacement of the ITs in preliminary analyses to the empirical descent of the ITs which was measured by comparing the non-weight-bearing and weight-bearing MRI scans [22]. When tested on the ACB cushion, a downwards displacement was prescribed on the superior surface of the IT so that the final distance between the skin and the base of the ACB cushion (the clearance above "bottom-out") was 32 mm, slightly above the 1-inch distance recommended for keeping away from bottoming-out [32]. The front and back planes of the buttocks and cushions were fixed in the perpendicular direction to assure thin slice model conditions. The inferior surface of the cushion was fixed for all translations and rotations, and tied interfaces were defined between all the tissue components. Frictional sliding was defined between the skin and the cushions with the coefficient of friction set to 0.4 in all simulations.

Meshing the model variants was performed using ScanIP® module of Simpleware® [30], with specific refinements to the skin and to the muscle and fat tissues near the tip of the ITs. Tetrahedral elements were assigned to the tissues, CFC and ACB cushion while hexahedral elements were assigned to the flat foam cushions. The FE simulations were all set up using PreView of FEBio, analyzed using the Pardiso linear solver of FEBio in its structural mechanics mode, and post-processed using PostView of FEBio [33].

Critical Characteristics of Effective Wheelchair Cushion Designs

Now that the etiology of PUs is better understood, the pathoanatomical variations in individuals have been considered, and the aforementioned computer simulation tools are available, five key characteristics of effective wheelchair cushion designs can be identified.

Immersion and Envelopment

The basic elements that, in combination, represent the potential cushioning performance of a cushion are immersion (the depth that a body penetrates into the surface), and envelopment (the intimacy of the cushion to the body). The ISO defines the characteristic of envelopment as "the ability of a cushion to conform, so to fit or mould around the irregular shape of the body" [34]. The importance of these characteristics are articulated by the ISO wheelchair cushion testing standards committee in the introduction of the Wheelchair Seating—Part 12: Envelopment testing technical specification, which provides detail of test equipment and method, for the measurement of "performance" of a wheelchair cushion intended to use immersion and envelopment to reduce local areas of pressure (by effectively supporting more tissue) [35].

In our biomechanical analysis, in order to quantify the extent of immersion and envelopment of the buttocks into the cushions a parameter α was defined, being the percentage of skin area that is in full contact with the cushion surface [27]. Stress concentrations formed in the gluteus muscle near the tip of the IT in both ACB and foam cushion simulations, but immersion was considerably greater in the case of an ACB cushion. Accordingly, α for the ACB cushion increased up to 91–93%, but only reached 58–65% for the foams. Consistent with the superior fit of the ACB cushion to the body contours, peak stress components in all tissues were four-orders-of-magnitude lower with the ACB cushion, with respect to the foams. We attribute this advantage of the ACB cushion in lowering peak tissue stresses to the substantially greater buttocks immersion it facilitates, which creates a much larger contact area for load transfer. As the contact area between the cushion and the skin increases, loads are transferred more uniformly, minimizing potentially hazardous areas of stress concentrations. Since sustained tissue loads imply safe sitting time for wheelchair users, as suggested in [36, 37], it appears that ACB cushions hold a benefit over foam cushions by providing greater immersion, which lowers internal tissue loads.

When we looked at the process of weight regaining on flat foam cushions, for example after completing a pushup, we found that skin shear loads exhibited a strongly non-linear increase, with a greater slope during the first half of the loading period, while fat shear loads increased linearly through the entire time course of loading [26]. This finding suggests that when sitting on standard foam cushions, the more sensitive period with respect to skin integrity is during initial skin-support contact. Since the skin-support contact area during the initial contact is relatively small, loads are temporarily more concentrated than when reaching full weight bearing. This phenomenon emphasizes the importance of large contact area for load transfer, as reflected in appropriate immersion and envelopment of the buttocks during sitting.

Adjustability to the Uniqueness of the Individual, at the Initial Fitting

Each and every user of a wheelchair cushion has morphologies, pathologies, and risks of tissue breakdown that are unique to them. While not all of the risks can be reduced, the cushion plays an important role in preventing PUs by having the capability of intimately adjusting to the individual, to achieve the desired immersion and envelopment to minimize the deformation that leads to internal tissue stresses and strains.

In our analysis, we introduced several particular risks, the first being the risk from previous PUs and the resulting scars. Investigating the influence of soft tissue scarring on peak tissue stresses during sitting or repositioning on flat foam cushions revealed that skin shear stresses increase in and around the (less deformable) scar. Importantly, the extent of the increase in loads within and adjacent to the scars strongly depended on the scar geometry, with the highest skin loads developed when a hypertrophic scar was present. This indicates that when sitting on a flat foam cushion, scarred skin is more vulnerable to a second breakdown event, especially if a thick (hypertrophic) scar has formed, which delineates new implications for the treatment of existing wounds to minimize hypertrophic scarring, and for risk assessment of individuals with a history of PUs [26]. Interestingly, when seated on an ACB cushion, soft tissue scarring induced, in general, lower peak stress values in the soft tissues of the buttocks with respect to the stress levels in the (non-scarred) reference case [28]. Peak effective and shear stresses in the skin decreased by up to 40% in all the simulated scar cases, while peak compressive and tensile stresses decreased by up to 41% in 8 out of 10 simulated scar cases. Likewise, peak stresses in fat tissue of scarred buttocks decreased on the ACB cushion with respect to the reference case by 40–65%, while gluteus muscle peak effective and shear stresses decreased by 10–45% in 9 of the 10 scars simulated. Furthermore, we tested the most severe soft tissue scar cases on a flat foam cushion, for direct comparison. We found that on a flat foam cushion, severe hour-glass shaped scar, involving muscle fat and skin tissues, causes an average increase of 155% and 70% in peak fat and muscle stresses, respectively, when compared against the reference, non-scarred, SCI anatomy on the same flat foam cushion. We concluded that the adjustability of the ACB cushion allows for improved stress distributions in the soft tissues of the buttocks, in the presence of scarred, stiffer tissue areas, compared to flat foam cushions.

We also considered the specific effects of bone adaptation (BA), muscle atrophy (MA), muscle spasms, and since an individual may experience all of these effects, the combination of all of the above were considered [27]. Sitting on the ACB cushion resulted in a substantially different loading state in the buttocks tissues with respect to flat foam cushions when all of these potential conditions were considered. Specifically, while BA increased peak stresses in muscles when seated on foams (8% increase in effective stress), the ACB cushion consistently promoted the opposite effect (41% decrease in effective stress). Likewise, though MA increased fat and skin effective stresses when sitting on the foams (by 57% and 37%, respectively), the ACB cushion was again able to reduce these stresses (by 60% and 23%, respectively) [27]. Therefore, the ACB cushion holds an additional advantage over foams when it comes to coping with and adjusting to the aforementioned chronic SCI pathoanatomies.

Adaptability to Movement and Activities of the Individual Throughout the Day

Our recent research [27] has demonstrated that a critical characteristic of effective wheelchair cushions is their ability to adjust to the body by providing adequate immersion and envelopment, which can greatly reduce tissue deformations, thereby preventing the tissue and cell damage associated with DTI. In Gefen [38], the remarkable disuse-related physiological changes that occur to the seated body over time have been reviewed [38]. Both papers point to a critical need for cushions to adjust to the body of the individual, both at the initial seating assessment as well as when changes occur over time. There exists another critical characteristic of wheelchair cushions however; the ability to adapt to changes in positioning associated with daily living, without the user having to actively adjust the cushion. While many cushions meet the first three characteristics we have described, there are very few which can also adapt to these activities of daily living without a conscious, active intervention by the individual. An example focusing on footwear follows. We studied two cushion technologies and five cushion variations. All cushions were code-verified in the US Medicare system as “adjustable skin protection cushions” and adjusted per manufacturer’s recommendations. Refinements were made using an XSensor pressure mapping system as in rehabilitation clinics, to achieve maximum contact area and minimize peak pressures (Fig. 2.2). A test subjects wearing tennis shoes was asked to sit in a Tilt-in-Space Chair set to 0°, 30° and 45° (without further adjustments to the cushions) and pressure maps were recorded. The footwear was then changed to four” high heels and data collection was repeated (Fig. 2.2). While inherent differences were observed across product performances, we found that in some products, changing the shoe type had a dramatic influence on peak pressures and contact areas for all postures. Measures of cushion efficacy in protecting users must suitably assess the way the cushion performs as the user’s body and function change over long times. Equally important however, is the cushion’s ability to automatically adapt to changes in the user’s sitting position throughout the day, which is influenced by numerous factors, including the wardrobe as shown here. It is unrealistic to expect that users would manually adjust their cushion each time they change clothes or perform an activity, so the self-accommodation of these common changes in positioning, and even wardrobe, is a critical characteristic of a good wheelchair cushion.

Adjustability to the Individual, Throughout the Subsequent Weeks, Months, and Years

Numerous wheelchair cushions are sufficiently adjustable to meet the needs of the individual, initially, to immerse and envelop the body. However, while many cushion succeed in this aspect, not all of them have the capability of accommodating all of the changes that may occur to the body during the useful life of the cushion. In our analysis, when simulating bodyweight changes which are typical to the first

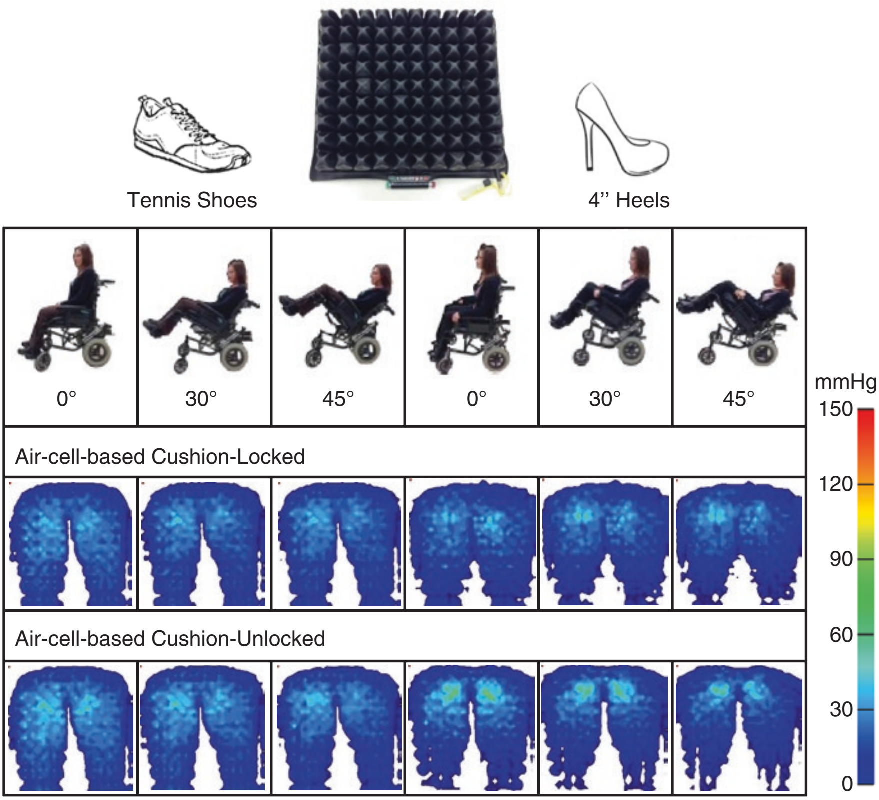

Fig. 2.2 Contact pressure maps of a test subject sitting on an air-cell-based (ACB) cushion, wearing either tennis shoes or four” heels, in a Tilt-in-Space chair set to 0°, 30° and 45°. This is one example of a superior performance of the ACB cushion technology, as it automatically adapts to changes in the user’s sitting position and is negligibly affected by wardrobe, both in a constricted airflow configuration (**Locked**) and in a free airflow configuration (**Unlocked**)

months and years after the acute injury (between −25% and + 40% fat mass) on CFCs, we found that effective and shear strains and stresses increased considerably with the chronological time-course of disuse [29]. For example, the peaks and respective ranges of effective and shear strains which developed in fat tissue increased by ~220% and 110%, respectively, in the model variant where the fat mass was increased by 40%, with respect to the ‘ideal fit’ reference model. In addition to that, the inhomogeneity in the strain and stress distributions in muscle tissue also increased with the simulated time-course post injury, resulting in greater strain and stress values in the model variants where the fat mass was increased [29]. FI and MA exacerbated the inhomogeneity of strain and stress distributions in both fat and muscle tissues and resulted in increased strain and stress values, particularly when severe weight-gain (additional 40% fat) was considered as well. For example, the peak effective and shear strains which developed in muscle tissue when a combination of severe weight-gain and severe FI was simulated increased to 24% and 15%, respectively, compared to strain values of 11% and 6% in the ‘ideal fit’ case. Combining these, we found that a CFC which has been fitted at a time close to the SCI but has not been replaced for several years thereafter substantially loses its efficacy in protecting patients from developing PUs, particularly DTIs, since shear loads and deformations are increasing internally in the soft tissues as the body responds to the chronic sitting and disuse [29]. Considering that within several months, at the latest, a SCI patient is expected to gain bodyweight and additional fat mass, extra-muscularly and intramuscularly, lose gluteal muscle mass and experience flattening of the ITs due to BA, the individual’s anatomy is, in fact, changing progressively and remarkably, but the CFC does not. As these changes take place and progress over time, the cushion’s contoured design quickly becomes irrelevant to the altered anatomy, both in terms of the adapted external buttock surfaces and the internal pathoanatomy, which can place patients at a considerable risk for PUs and DTI. Therefore, our simulations results highlights the importance of sufficient ‘adjustability’ over time, to maintain ideal immersion and envelopment, as a critical design feature for any cushion that is meant to protect against PUs.

Durability, Over Time, Throughout the Weeks, Months, and Years

We have demonstrated the need of the cushion to adapt to the individual over time to maximize immersion and envelopment; however, in order to maintain the level of efficacy, the cushion must not only be adjustable, but also durable. Recently, Sprigle and Delaune [39] published their research comparing the performances of foam cushions, gel-filled cushions, and air cushions [39]. Their testing revealed that foam cushions have a much higher prevalence of fatigue, as compared to air-filled cushions and gel-filled cushions. Twenty-one percent of foam cushions, 20% of gel-filled cushions, and 16% of air-filled cushions showed visible damage throughout the duration of the study. Furthermore, they showed that foam cushions used 12 h per day for longer than a year were 7 times more likely to have tears, 2.23 times more likely to be discolored, and 3 times more likely to show brittleness than those foam cushions used for under 12 months. In Ferguson-Pell [40], the fact that foam cushions deteriorate with time, even when they are not being used, is highlighted [40]. This deterioration is caused by the brittleness of the polymer matrix, leading to fractures and softening of the product. Soiling and liquid products will further break down the foam, as they are particularly susceptible to moisture. The longer the foam cushion is in use, the greater and the faster the damage will progress. Gel cushions also show age, often times developing hard or consolidated regions that need to be kneaded to break up and to help prolong the lifespan of the product.

The ISO wheelchair seating working group (ISO/TC 173/SC 1/WG11) has recognized the importance of performing bench testing of wheelchair cushions, while simulating aging of the products, to capture the deterioration in efficacy [34]. The standard ISO 16840–6 “Wheelchair Seating—Part 6; Determination of the changes in properties following simulated extended use—seat cushions” introduces this critical need to evaluate cushion performance over time [41]. First, the cushion is tested to characterize the properties of a new cushion, then it is subjected to multiple simulated aging processes, and finally re-tested to characterize the changes in properties. The suggested aging processes include thermal accelerated aging, bacterial, urinary and faecal soiling, disinfection, laundering, and ultraviolet and ozone exposure, all of which can be expected during the life of the cushion.

The Impact of Research on Industry, Regulation and Reimbursement Policies

Science and public policy are in a virtual “tug-o-war” regarding beneficiary access to the goods and services that address their needs. When credible science exists then policy makers are compelled to take notice and will find it difficult to ignore in establishing coverage and payment policies. However, when scientific knowledge is insufficient, and this may still be the case in PU prevention and treatment research, policymakers are prone to establishing coverage and payment rules that primarily focus on financial objectives, or are biased towards broad characterization and commoditizing of medical equipment, with less attention to ensuring that products are indeed capable of meeting the individual’s medical needs. The problems that this creates are exacerbated by the fact that health care policies, coverage and payment are often being compartmentalized by care settings with no consideration of the individual’s care and treatment throughout the continuum of care. Over time, this may actually increase the overall costs to the individual and the healthcare system, as the individual’s needs are unmet and further damage occurs. For example, if certain wheelchair cushions that are prescribed and reimbursed for prevention or care of PUs do not actually provide the intended benefits to the individual (though policy makers assumed they would, due to a gap in understanding), the prevalence and incidence of PUs in the wheelchair user population will actually rise, thus pushing the healthcare costs upwards.

Initial steps in bridging this gap were taken in the early 2000s in the US, when the Medicare system adopted a method of evaluating the depth of supported immersion a cushion which is considered a “skin protection cushion” could provide, specifically HCPCS (Healthcare Common Procedure Coding System) codes E2603, E2604 (“nonadjustable skin protection seat cushions”), and E2622, E2623 (“adjustable skin protection seat cushions”). An analog of the pelvis, proposed by Springle et al. [42], was developed for bench testing, constructed of two inner cylinders, which represent the position of the ischial tuberosities, and two outer cylinders, 40 mm higher, which represent the relative positions of the greater trochanters [42] (Fig. 2.3). The test method of using a loaded contour jig is required by the Durable Medical Equipment Pricing Data Analysis and Coding (DME PDAC), a Medicare contractor who evaluates cushions and classifies them by Medicare HCPCS codes (https://www.dmepdac.com/).This coding determines which cushions can be prescribed to individuals, with costs reimbursed by the Medicare system. It is therefore a critical function intended to ensure patient access to appropriate medical equipment.

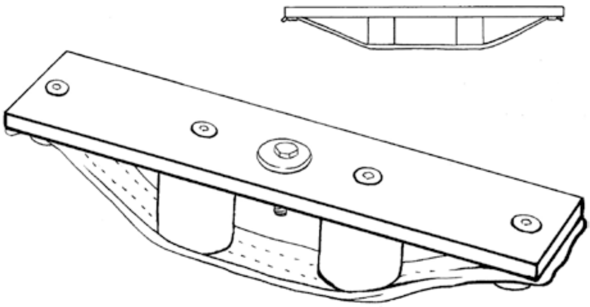

Fig. 2.3 An analog of the pelvis, proposed by Springle et al. [42], constructed of two inner cylinders, which represent the position of the ischial tuberosities, and two outer cylinders, 40 mm higher, which represent the relative positions of the greater trochanters

The loaded contour test does provide a rudimentary method of evaluating whether a cushion may have the depth capacity to accommodate the immersion of the pelvis in a standard and overload condition, and it does serve as a simple threshold that can eliminate cushions from making a skin protection claim if they do not pass this test. However, a cushion could easily be designed for the sole purpose of passing this specific test, without providing clinical benefit. Additionally, the standard requires the cushion to be re-tested after simulated aging of 18 months, but no simulated aging techniques are recommended. With the publication of ISO 16840–6, the hope is that this new wheelchair cushion aging standard will be adopted by the Medicare system in part or in whole to provide definition of how the cushions should be aged. Furthermore, the term "adjustable" is applied to the cushion codes E2622 and E2623, which must meet all of the requirements of the non-adjustable cushions.

Summary and Conclusions

There is a growing understanding that appropriate immersion and envelopment of the body into a wheelchair cushion are key factors in protecting patients against sitting acquired PUs, as they allow for improved dissipation of soft tissue loads and deformations. A good wheelchair cushion should first be capable of accommodating the seated buttocks, providing the adequate immersion and envelopment of the seated individual, during the initial seating assessment. This feature was acknowledged when the Medicare system adopted a method of evaluating the depth of supported immersion a cushion provides to determine whether it should be classified as a "skin protection cushion". A good wheelchair cushion should be able to conform to individuals with different anatomies, or sometimes disuse-related pathoanatomies, and offer the optimal biomechanical conditions in the soft weight-bearing tissues of the buttocks. Then, the cushion should be able to maintain its efficacy over the time of intended use. It should also be able to accommodate changes in posture and weight shifts associated with daily living, and conform to the remarkable disuse-related anatomical and physiological changes, which are expected in the months and years following a SCI. Furthermore, the cushion should maintain its physical and mechanical properties as well as its performance over time and despite exposure to various degenerating conditions, which can be expected during the life of the cushion.

In this chapter, we demonstrate the novel use of FE computational modeling in wheelchair cushion assessment and its ability to isolate different risk factors associated with either the cushion or the individual. Given the advances in understanding that tissue deformation is a key contributor to DTI and PUs, and the availability of new tools to assess relative protection against deformation through immersion and envelopment, during everyday life, over time as the individual changes, and over time as the cushion changes, there is a considerable gap between public policy and the tests which are currently applied to evaluate the efficacy of cushions, and the challenges and measures that should be applied.

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