Ab Initio Calculations of Seat Interface Pressures

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ABSTRACT

In this project we use a finite element model to investigate seat interface and interior seating pressures in spinal cord patients who are confined to wheelchairs. Sustained high pressures under the bony prominences and the associated loss of circulation are the primary mechanisms in the development of pressure ulcers. Previously a rudimentary finite element model of the buttocks and lower pelvis was developed to study these seating pressures. The goal of this project is to refine the existing model by using a more realistic model of the pelvis, which should distribute the weight of the upper torso more accurately. The predictions of this model will be compared with actual laboratory data obtained from a Novell Pliance pressure-sensitive mat.

INTRODUCTION

Patients who suffer severe spinal cord injuries and are confined to a wheelchair often develop pressure ulcers[1]. Pressure ulcers unquestionably result from a very complex set of risk factors, but the primary mechanism is sustained high pressure and the associated loss of circulation[2]. The highest pressure occurs beneath the ischial tuberosities (IT) due to the concentration of localized forces[3]. The relatively recent advances in multiple sensor pressure sensitive mats with matrices of sensors have made it possible to measure seat interface pressures[4]. However, it is not clear whether the interface pressure provides enough information, since there is evidence that the highest pressures are interior[5] and that pressure ulcers develop internally and spread toward the surface.

Mechanical and computer models of the buttocks and lower pelvis have been used to study the interior pressures that occur in a seated human. Recently, Bidar *et al* [5] used a rudimentary finite element model of the buttocks and lower pelvis to study the effect of cushion thickness on interior pressures. The weight of the upper torso was divided into two loads; one applied to the top surface of the IT, and the other as a uniform pressure on the top surface of the buttock. Although the model was simple, it could be adjusted to fit interface pressure readings obtained with a pressure sensor mat. The only fitting parameter was the relative fraction of the upper torso load applied to the upper surface of the buttock and the IT. Surprisingly, the loads on the ITs were found to be a small fraction of the torso weight (15%), with the majority of the weight supported by the buttocks. This disagrees with previous studies, where it was assumed that the total upper torso weight should be applied to the IT[6,7].

In order to determine the relative fraction of the weight of the upper torso that is transmitted to the buttocks by the surfaces of the pelvis other than IT, an *ab initio* model with a more realistic pelvis and no adjustable parameters has been developed.

METHODS

The current model is a refinement of the Bidar et al model and is similar to the mechanical model used by Reddy et al.[8] See Bidar et al[5] for details. The Bidar et al model considered each buttock as a horizontal axisymetric slab with a radius of 8 cm and width 8 cm with an IT situated along the axis. In the current study, a circular bony plate situated on the upper surface of the buttocks simulates the surfaces of the pelvis, other than the IT see Figure 1. These distances were chosen as effective approximate distances taken from an adult skeleton. The weight of the upper torso is applied uniformly on the upper surface of the pelvis. The buttocks were considered to consist of nearly incompressible "soft" human tissue with a Young's modulus of 47 kPa and a Poisson's ratio of 0.49 [7]. The Young's modulus of the bone was 10^4 kPa and 0.31 was used for the Poisson's ratio. The minimum distance between the point of the IT and the skin was 1.5 cm before loading. The weight of the upper torso was taken to be 65% of the total body weight.

RESULTS

The computational model predicted that within the soft tissue, the area of high stress was concentrated within a centimeter or two of the IT with the maximum compressive stress just below the surface of the IT, as shown in Figure 2. The seat interface pressures from the model were compared with those taken from pressure mat obtained from a 70 kg single subject sitting in a standardized position on a hard surface was used, as shown in Figure 3. The pressure was put into radial form by measuring the average pressure versus radial distance from the axis of the IT for each buttock. From the model we can find the pressures at the top surface of the model defined by Bidar et al, which are shown in Figure 4. From our results we calculated that the total load supported by the IT is only 20% of the upper torso weight, which supports the loading assumptions used in their model.

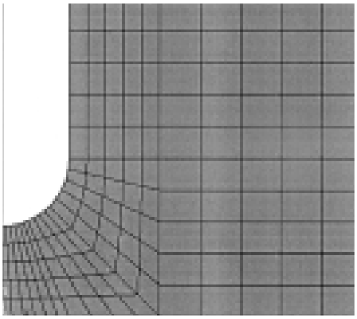

Figure 1. Cross section of the axisymmetric pelvis model. White represents bone, while grey represents the soft tissue of the buttock.

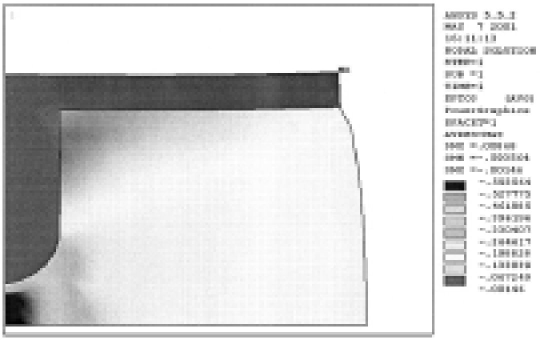

Figure 2. Contour plot of 3rd principal stress. The blue represents maximum compression while red represents minimum compression. This is the predicted compression, made in ANSYS® finite element software.

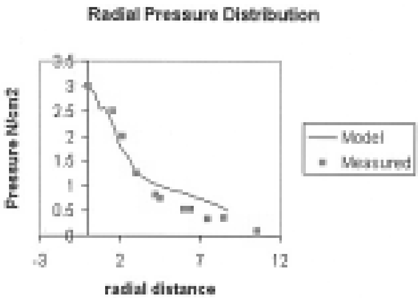

Figure 3. Comparison of predicted and clinically observed seat interface pressures

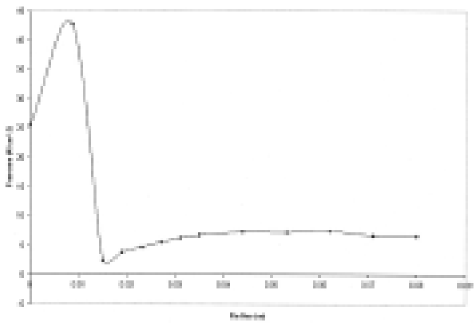

Figure 4. Pressure 4cm from the interface versus radial distance from the axis of the IT. This corresponds to the top boundary of the Bidar et al model.

DISCUSSION

The current model fixes many of the flaws of the initial model. However, there are still some modifications that could further enhance upon the current model. A precise computer model of the human pelvis is impossible at this point due to limitations in our computational power and restricted by the number of nodes in our version of ANSYS®, however some further modifications are possible. Two of these would include rounding of the bottom surface of our buttock model and making a pelvis with angled surfaces.

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