Detection of Pressure-Related Deep Tissue Injury in Permanent Wheelchair Users using Real-Time Hertzian Contact Analysis

[1]Limor Agam and [1]Amit Gefen
[1]Department of Biomedical Engineering, Faculty of Engineering, Tel Aviv University, Israel
Web: http://www.eng.tau.ac.il/~msbm/, email: gefen@eng.tau.ac.il

INTRODUCTION

Pressure ulcers are one of the most serious complications in paralyzed and immobilized patients, with severe consequences such as sepsis, osteomyelitis, renal failure and myocardial infraction. It is hypothesized that the most injurious type of pressure ulcer, called deep tissue injury (DTI) first occurs under the IT, where compression stress concentrations in paraplegics can be as high as 101±38 kPa (average ± standard deviation) according to a recent integrated MRI and FE study of sitting subjects []. The wound then spreads with a snowball effect [,], as tissue undergoes structural and mechanical changes, and draws more stress to the injured area. This indicates that constant, immediate estimation of stresses under the IT is critical for assessment of injury initiation – once it occurs, damage is not only irreversible, but probably very difficult to arrest.

Previous studies have monitored contact stresses in real-time under the buttocks, but these have been proven to be much milder from the situation in the muscle itself and especially near the IT []. Other studies have adopted a much more accurate approach, estimating stresses in the muscles under the IT using finite element (FE) and complex calculations, at the cost of immediacy and portability [,,]. Never before have the two approaches been fused together in order to monitor stresses inside the muscle in real-time, while allowing the patient mobility.

AIMS AND OBJECTIVES

The goals of this study were (i) to develop a system for real-time evaluation of stresses in the soft tissues under the IT, and (ii) to develop a light and portable apparatus for home use that would serve as an alarm system, warning a paralyzed sitter or his/her caretakers that internal tissue stresses have exceeded a recommended limit.

METHODS

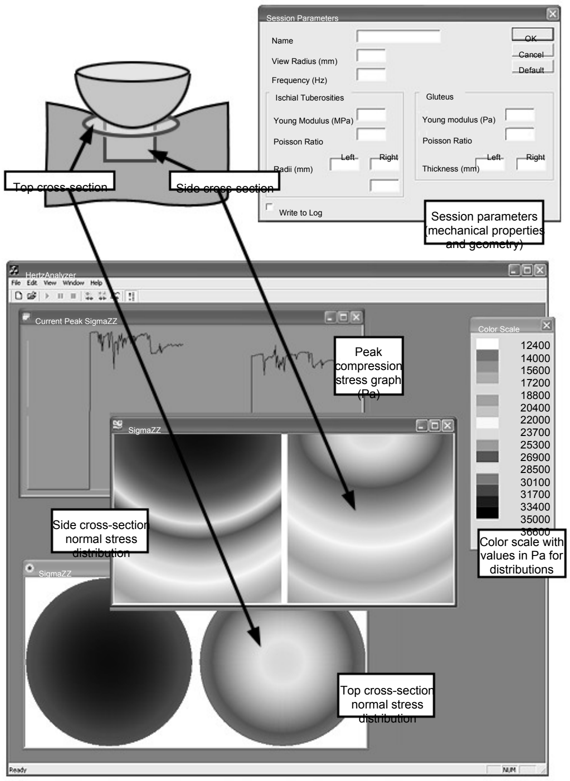

Figure 1: The Hertz model software: parameters dialog, cross-sections of left and right gluteus, peak maximal compression stress graph, color scale.

A software simulating the IT-muscle contact problem was developed (Visual C++ 6, Microsoft Co., Redmond, WA, USA), employing the classical Hertz sphere (IT)/half-space (soft tissue) contact model. Real-time calculation and depiction of gluteal muscle stress distributions in both top and side cross-sections is available (Figure 1), as well as graphs of peak stresses, the scale of which is constantly updated as stresses progress. An interface allows the user to input individual parameters such as the subject's bone curvature and tissue thickness, based on X-ray or MRI scans, as well as the bone and muscle's elastic moduli and Poisson's ratios. The radius of view around the points of interest, and frequency of sampling may be adjusted. The color scale is automatically adjusted to the patient's range of stresses during the session. Data may be saved in a log file, for more complex analysis at a later stage.

Experiments on eight subjects were conducted using a wheelchair and an air cell cushion which was adjusted to each individual (Figure 2). Six healthy subjects were monitored for at least 80 minutes while watching a movie of their choice. Two paraplegic subjects were monitored for two to three minutes. Interface pressures were measured at a rate of 1 Hz, using a pressure mat which was placed in between sitter and cushion (256 2.5X2.5 cm2 sensors, Tactilus®, Sensor Products LLC).

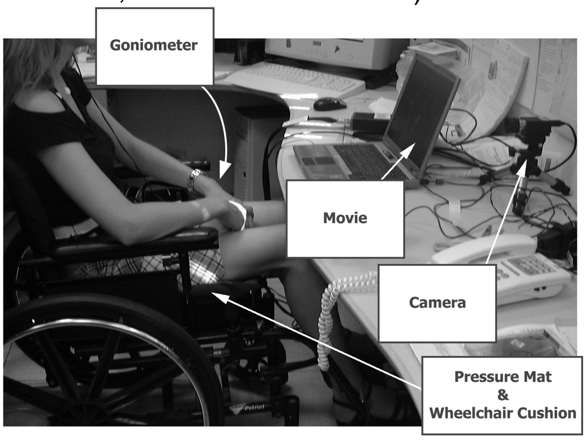

Figure 2: Experimental setup

RESULTS

Comparison of maximal compression stresses, as recorded during sitting sessions of the same six normal subjects (left and right sides of buttocks pooled) by the FE method [] and the current Hertz model, showed that the Hertz model is significantly more severe in estimating compression stresses (p=0.01), resulting in an average that is about 1.43 times that of the FE. The average-difference comparison method [] showed that all stress data points were within the 2SD lines.

The mean maximal peak compression of all healthy subjects during 60 minutes of sitting was 49 6.6 kPa. The mean maximal peak compression stress for healthy subjects during two minutes was 46 6 kPa, while in the paraplegic group a mean of 129 66 kPa was recorded. The mean peak stress of the paraplegic subjects is 2.8 fold of that recorded in the healthy ones.

Mean dose of peak compression stress over two minutes for healthy subjects was 5146 718 kPa·sec on z-axis, and 10053 1446 kPa·sec on r-axis. Paraplegic subjects were characterized by a mean dose of 14652 7734 kPa·sec on the z-axis and 27964 14384 kPa·sec on the r-axis. Paraplegic doses on both axes were 2.8 fold those of healthy subjects.

DISCUSSION

The new method examined in this study gives a reasonable estimate of soft tissues stresses underneath the IT. It is not as accurate as using FE, as it gives an overestimation that is about 1.43 fold that of the FE model for normal subjects, but given the practical employment possibilities of the software for uses such as described in the aims above, we believe a slight overestimation is safer than underestimating. Stresses of paraplegic subjects were also overestimated, but a sounder conclusion may be reached when more data has been acquired and the FE studies in our group completed.

CONCLUSION

We conclude that the new method is suitable for estimating the stresses under the IT, although it is not accurate. Since it is more severe in its evaluation than the FE, patients using this apparatus in every day life may be prompted to move more often than they need to, but no harm is known to come from that.

REFERENCES

[1.] Salford, AL. Proceedings of BLLHDR 2005. Salford, UK,.2005
[2.] Salford AL, Salford, AL, Salford AL et al. J. Biomech., 40, 1881-1890, 2006
3. Salford AL, Lower limb walking.
University Press, Salford, UK, 2002
Seidel GK, Marchinda DM, Dijkers M. et al. J. Biomech, 28: 8; 995-998, 1995.

[1.] Linder-Ganz E, Shabshin N, Itzchak Y, et al. Proceedings of the ASME 2007. Keystone, Colorado, USA, 2007.
[2.] Gefen A, Gefen N, Linder-Ganz E, et al. Journal of Biomechanical Engineering, 127:3; 512-524, 2005.
[3.] Kuroda S, Akimoto M. Journal of Nippon Medical School, 72; 174-178, 2005.

4. Le KM, Madsen BL, Barth PW, et al. Plastic and Reconstructive Surgery, 74:6; 745-756, 1984.
5. Breuls RGM, Sengers BG, Oomens CWJ, et al. Journal of Biomechanical Engineering, 124:2; 198-207, 2002.
6. Breuls RGM, Bouten CVC, Oomens CWJ, et al. Journal of Biomechanical Engineering, 125:6; 902-909, 2003.
7. Linder-Ganz E, Shabshin N, Itzchak Y, et al. Journal of Biomechanics, Aug 2006.
8. Bland JM, Altman DG. Lancet, I; 307-310, 1986.