XXIV CONGRESS OF THE INTERNATIONAL SOCIETY OF BIOMECHANICS
XV BRAZILIAN CONGRESS OF BIOMECHANICS

THE EFFECT OF KYPHOPLASY ON THE LOAD TRANSFER WITHIN THE LUMBAR SPINE CONSIDERING THE BIOMECHANICAL RESPONSE OF VERTEBRAL BODIES AND LIGAMENTOUS TISSUE

[1]Alexander Tsouknidas, [2]Savvas Savvakis, [2]Nikolaos Tsirelis, [3]Kleovoulos Anagnostidis and [1]Nikolaos Michailidis
[1]Department of Mechanical Engineering, Aristotele University Thessaloniki, 54124 Thessaloniki, Greece
[2]BETA CAE Systems S.A., 54005 Thessaloniki, Greece
[3]Cardiff Spine Unit, University Hospital of Wales ,University Hospital Llandough, United Kingdom

INTRODUCTION

Vertebral compression fractures (VCFs) caused by osteoporosis are an increasingly common occurrence. The annual incidents of vertebral fractures in Europe among elderly (50–79) are estimated at 1.1% for women and 0.6% for men, while epidemiologic studies foresee an increase of these numbers in the future [1]. Approximately 85% of these fractures are due to primary osteoporosis and the remainder due to secondary osteoporosis or malignancies [2]. These VCFs lead to progressive sagittal spinal deformity and changes in spine biomechanics. Patients diagnosed with a prevalent vertebral fracture are susceptible to further trauma in adjacent spine levels, with a fivefold increased risk [3,4].

Balloon kyphoplasty is a minimally invasive surgical treatment for osteoporotic and osteolytic vertebral compression fractures with promising clinical potential, during which a filler material is percutaneously injected into a cavity of a degenerated vertebral body, created by an inflatable tamp. Next to reversing kyphosis, cemented augmentation also results in high local rigidity within a Functional Spine Unite (FSU) and retrospective clinical studies have indicated new VCFs as a potential late sequela of the reinforcement procedures [5]. It remains however elusive whether this is the etiology [4,6] or a symptomatic condition of the gradual loss of bone mineral density due to evolving osteoporosis [7,8].

The pathogenesis of fractures at adjacent non-treated spine levels has been heuristically investigated both in vitro [9,10] and in vivo [11]. Experimental studies are however conducted on FSU's originating from different spine levels, age groups and varying surgical approaches and have thus been indicated as methodologically flawed [6]. This hinders a collective evaluation of the existing literature, as different hypothesis and conclusions render it unclear whether these trends will hold true once deducted to other patients. Finite Element Analysis (FEA) has been used to determine the in situ effect of cemented augmentation on the load transfer within a FSU [12] indicating increased pressure in the intervertebral disc and deflection of the vicinal endplate, that could provoke subsequent fractures. The biomechanical alterations of ligaments however could not be reflected, as these were simulated by cable elements, capable of enduring tension only. Recent FEA continue to focus on the response of the adjacent vertebra considering motion segments of 3-5 vertebral bodies [13,14] with ligamentous tissue either modeled by two nodal elements or neglected at all.

In this investigation a FEA of a bio-realistic lumbar (L1-L5) spine is introduced to compare the biomechanical response of its preoperative state to the postsurgical cemented augmentation, both for bony and connecting soft tissue. This approach is based on the preliminary hypotheses that cement injection exaggerates force transmission to the adjacent vertebral bodies, thereby predisposing those levels to future fractures. The effect of uni- and bipedicular filling with Polymethyl-methacrylate (PMMA) was examined for loads encountered in diurnal activities.

METHODS

A lumbar spine was scanned in its entirety by high resolution Computed Tomography (CT). Upon reconstruction of the vertebral bodies, the intervertebral discs (IVD) and connecting ligaments were reverse engineered based on the surface of the interposing vertebrae [15]. Ellipsoid cavities were inducted into the L3 vertebrae to examine load transition over two adjacent levels towards L1 and L5 respectively. The cavity in the case of unipendicular kyphoplasty was filled by approximately 3ml of PMMA whereas in the scenario mimicking bipedicular reinforcement, two symmetric cavities of similar dimensions were introduced as shown in figure 1.

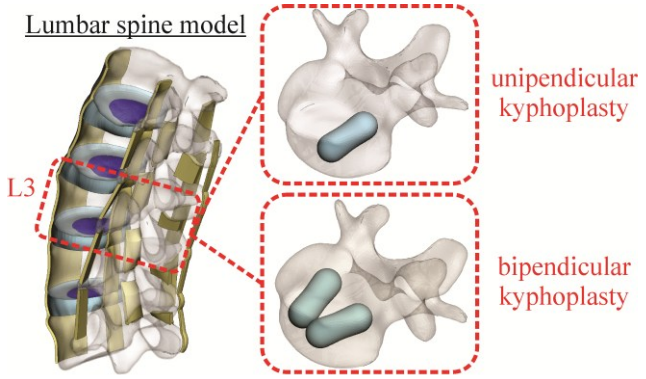

Figure 1: Uni- and bipedicular cavities introduced in the L3 vertebrae

The mesh grid was generated in ANSA (by BETA CAE Systems S.A.), in order to consider anatomic characteristics (i.e. integration of annulus collagen fibres). Convergence studies indicated the optimum mesh density in terms of processing time and results accuracy. To avoid element shear locking and hourglassing, hexa- and tetra-hedral second order elements with reduced integration were employed for all model entities consisting at least 4 element layers for ligamentous tissue. AnyBody was used to determine muscle activation on the Functional Spine Unit (FSU) during a mild running scenario (stance phase). The vertical force component of the ground reaction was registered and the time varying force profile considered along with vertical motion of the spine (as registered by AnyGait), to account for inertia phenomena. The model was simulated in Abaqus with non-linear, stress strain dependent material properties, thus allowing insight to occurring dynamic response of the FSU.

RESULTS AND DISCUSSION

Accumulated stress in the bone-PMMA interface of the treated vertebrae, suggest a degeneration of its structural characteristics compared to native specimens. This coheres to previous findings [16] as uni- and bipedicular filling exhibit varying stress fields, suggesting that the biomechanical response of the FSU depends not only on filler material but also on the injected volume and distribution [17]. The stiffness characteristics of a healthy vertebra would however compare favorably to a reinforced specimen [18] in either on scenario, mainly due to the capacity of an intact vertebra to distribute developing stress over its entire volume.

Both uni- and bipedicular filling exhibited encouraging restoration of intradiscal pressure with slightly increased values - pronounced stress distributions for the adjacent endplates (figure 2). This lead to deflection of the adjacent endplate, a phenomenon being more pronounced for bipedicular augmentation, a tendency also demonstrated by Politkeit et al. [12]. Both procedures indicated significantly heightened stress transfer towards the first adjacent vertebral level, gradually decreasing thereafter.

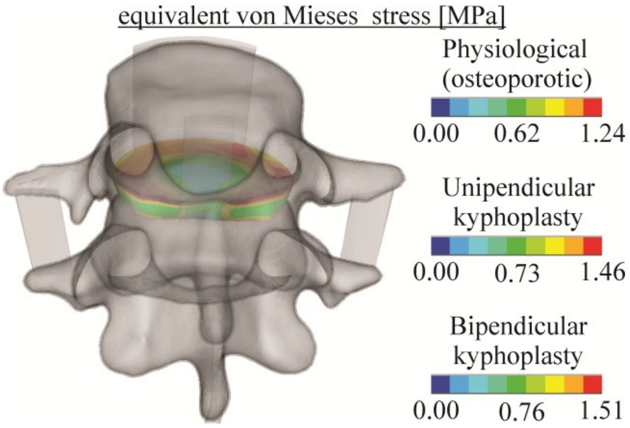

Figure 2: Stress distribution within an endplate adjacent to the treated vertebra.

Hyperphysiological stress values were also observed in ligamentous tissue connected to the reinforced vertebrae. These stress concentrations can lead to degeneration of the ligaments, increasing the range of motion within a spine unit [19], thus fostering even higher IVD pressures provoking increased load transfer to non-treated levels.

CONCLUSIONS

The introduced model was validated against literature data [20] for 3 static loading scenarios (flexion, bending and torsion). Based on the computed results an overall increase in the load transfer was observed in both treated models.

Stating however, that kyphoplasty increases the risk of fragility fractures in adjacent nontreated spine levels would be an oversimplification and fairly confusing statement. The reinforced postsurgical model studied in this investigation, like in most FE approaches [12,13,14], is compared to an intact spine segment rather to a fractured one. This does not facilitate the quantification of the negative effect of post traumatic kyphosis on the load transfer within a traumatized spine, which is worse by definition.

In retrospect, kyphoplasty maybe beneficiary to the overall load transfer within a spine segment, the localized growth in rigidity however, induced by cemented augmentation, is likely to contribute to subsequent collapse of adjacent vertebrae.

ACKNOWLEDGEMENTS

The Authors would like to acknowledge that this investigation was partially funded by the General Secretariat for Research and Technology of Greece under grant PE8(3227).

REFERENCES

1. European Prospective Osteoporosis Study (EPOS*), J Bone Miner Res.* **17**:716–724, 2002.
2. Cooper C, et al., J Bone Miner Res. **7**:221–227, 1992.
3. Heaney RP, *Bone*. **13(S2):** 23-26, 1992.
4. Lindsay R, et al., *JAMA*. **285(3)**:320-323, 2001.
5. Fribourg D, et al., *Spine*. **29**:2270–2276, 2004.
6. Hardouin et al., *Joint Bone Spine.* **68**:216-221, 2001.
7. Uppin AA, et al., *Radiology*. **226(1):**119-124, 2003.
8. Grados F, et al., *Rheumatology*. **39(12)**:1410-1414, 2000.
9. Boger A, et al., *Eur Spine J.* **16(12):**2118-2125, 2007.
10. Berlemann U, et al., *J Bone Joint Surg Br.* **84(5)**:748-52, 2002.
11. Fournol M, et al., *J Radiol.* **88(6)**:877-880, 2007.
12. Politkeit A, et al., *Spine*. **28(10)**: 991-996, 2003.
13. Villarraga ML, et al., *J Spinal Disord Tech.* **18(1):** 84-91, 2005.
14. Rohlermann et al., *Eur Spine J.* **15**: 1255-1264, 2006.
15. Tsouknidas A, et al, *J Appl Biomech.* **28(4)**:448-456, 2012.
16. Ferguson S, et al., Proceedings of the 47th Annual Meeting ORS, San Francisco, USA, Proceedings p. 280, 2001.
17. Chevalier Y, et al., *Spine*. **33(16)**:1722–1730, 2008.
18. Wilson et al., *Spine*. **25**:158-165, 2000.
19. Xiao et al., *Theor Appl*. **1**:064001, 2011.
20. Panjabi et al., *J Bone Joint Surg.* **76**:413–24, 1994.