PEDIATRIC CERVICAL SPINE BIOMECHANICS USING FINITE ELEMENT MODELS

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ABSTRACT

This study was conducted to determine the external and internal biomechanical responses of the one, three and six year old pediatric cervical spine structures under varying severities of combined compression-flexion and compression-extension, and axial tension load vectors. Our existing one, three and six year old pediatric lower cervical spine finite element models were used. The models in the three age groups were developed by incorporating the characteristic developmental pediatric anatomy such as the variations in the ossification patterns. The following components were included: vertebral centrum, neural arches, neuro-central and costal cartilages, growth plates, nucleus pulposus, anulus fibers, ground substance and all major ligaments. The three-dimensional models were exercised using the principles of geometrical and material nonlinear analyses. The resulting external and internal biomechanical responses were expressed in terms of the overall flexibilities; the intrinsic stresses in the bone and intervertebral anulus, and pressures in the nucleus and facet joints were determined under each load vector. All these output were compared with respect to the adult human spine. In general, decreases in the flexibilities were apparent with increasing age; this was independent of the load vector. However, the magnitudes of the decrease were nonuniform with respect to the age group, and the type and severity of the external load vector. Under tension, the intrinsic pressures in the joints decreased. Under combined compression-flexion and compression-extension loading, facet joint pressures were higher than disc pressures. The relative magnitudes of the pressures were dependent on the severity of the combined load vector. The anulus stresses, in general, were highest for the adult spine with the maximum magnitudes occurring at the most severe compression-flexion and compression-extension load vectors. The stresses generally decreased from the adult spine to the one year old pediatric spine. These characteristic differences with respect to the age and loading mode may assist to explain the intrinsic load sharing among the developing pediatric components and provide a likely explanation for the mechanisms of injury.

MATHEMATICAL MODELING of the human cervical spine is necessary to understand the extrinsic and intrinsic biomechanical characteristics under external loading conditions (Yoganandan et al. 1987; de Jager 1993). The stress analysis approach should be followed in order to determine these characteristics. Because of the geometrical complexity and material inhomogenity of the cervical spine structures, traditional continuum mechanics approaches cannot be used. Computer driven numerical finite element techniques offer a unique methodology to accurately simulate the three-dimensional hard and soft tissue geometry, incorporate detailed material properties of the various cervical spine components, and conduct parametric analyses so that the external and internal responses can be delineated.

Although the finite element method was introduced in 1956 to analyze structural mechanics problems, applications of this technique in the musculoskeletal biomechanics area started with the thorax modeling in 1970 and human vertebral column modeling in 1973 (Yoganandan et al. 1987). Since this period, finite element models of the human spine have proliferated. Efforts have included modeling of the isolated components such as the cancellous core of the lumbar vertebrae to the complex modeling of the segmented units of the spinal column. Continued advancements in numerical techniques, computer technology, and improvements in the pre and post processing software have made the finite element method a versatile tool for biomechanics applications. Research has been undertaken to investigate the stress analysis of the human low back under normal, injured and stabilized conditions, and to study the human cervical spine biomechanics. Of necessity, all these developments have been primarily limited to the adult human structure (Saito et al. 1991; Kleinberger 1993; Bozic et al. 1994; Dauvilliers et al. 1994; Teo et al. 1994; Yoganandan et al. 1995; Yoganandan et al. 1996; Kumaresan et al. 1997; Kumaresan et al. 1997; Voo et al. 1997; Yoganandan et al. 1997; Kumaresan et al. 1998). Only recently studies have begun to delineate the biomechanical responses of the pediatric spinal structures (Kumaresan et al. 1997; Kumaresan et al. 1998).

It is common knowledge that the pediatric cervical spine is not simply a scaled-down adult spine (Burdi et al. 1969). Considerable evidence exists in the anatomical and physiological literature that the growth and development process of the pediatric human affects the biomechanical responses including the injury patterns to this population (Gooding and Neuhauser 1965; Bonadio 1993; Bonadio 1993). For example, spinal cord injuries without radiographic abnormalities, commonly termed as SCIWORA, do not occur in the mature adult neck (Pang and Wilberger 1989). Despite these clinical identifications of the differences between the adult and pediatric cervical spine structures, and the epidemiological evidences relating the differing patterns of injury between the adult and pediatric human, relatively, there is a paucity of data to clearly delineate the external and internal biomechanical responses of the pediatric structures as a function of age under physiologically relevant external load vectors.

Consequently, our group has advanced efforts in the past to understand the biomechanical behaviors of the pediatric cervical spine structures. Our previous studies have resulted in the estimation of the biomechanical responses of a "representative" pediatric cervical spine using the principles of linear elastic analysis under physiologic pure moment-type flexion, extension, lateral bending and axial torsion force vectors (Kumaresan et al. 1998). Realizing the limitations of the linear model and the inability to characterize the responses as a function of age using a single representative model, further efforts were advanced to systematically create the one, three and six year old pediatric cervical spine structures. These age-specific nonlinear finite element models were exercised under pure force vectors: compression, flexion and extension. The external responses, i.e., compressive, flexion and extension stiffnesses of the pediatric structures were compared with that of the adult; these results are published (Kumaresan et al. 1997).

As a continuation, the present objective was to investigate the external and internal biomechanical responses of the one, three and six year old pediatric cervical structures under axial tension and combined load vectors. Compression-flexion and compression-extension loads at varying eccentricities were applied to all finite element models. The internal biomechanical responses were quantified in terms of the stresses in the vertebral body and intervertebral discs, and the pressures inside the intervertebral nucleus and facet joints. In addition, the external responses were defined in terms of the flexibilities under each load vector. All these biomechanical parameters were compared with the adult human cervical spine structure. The rationale for choosing these combined load vectors is provided later in the paper.

MATERIAL AND METHODS

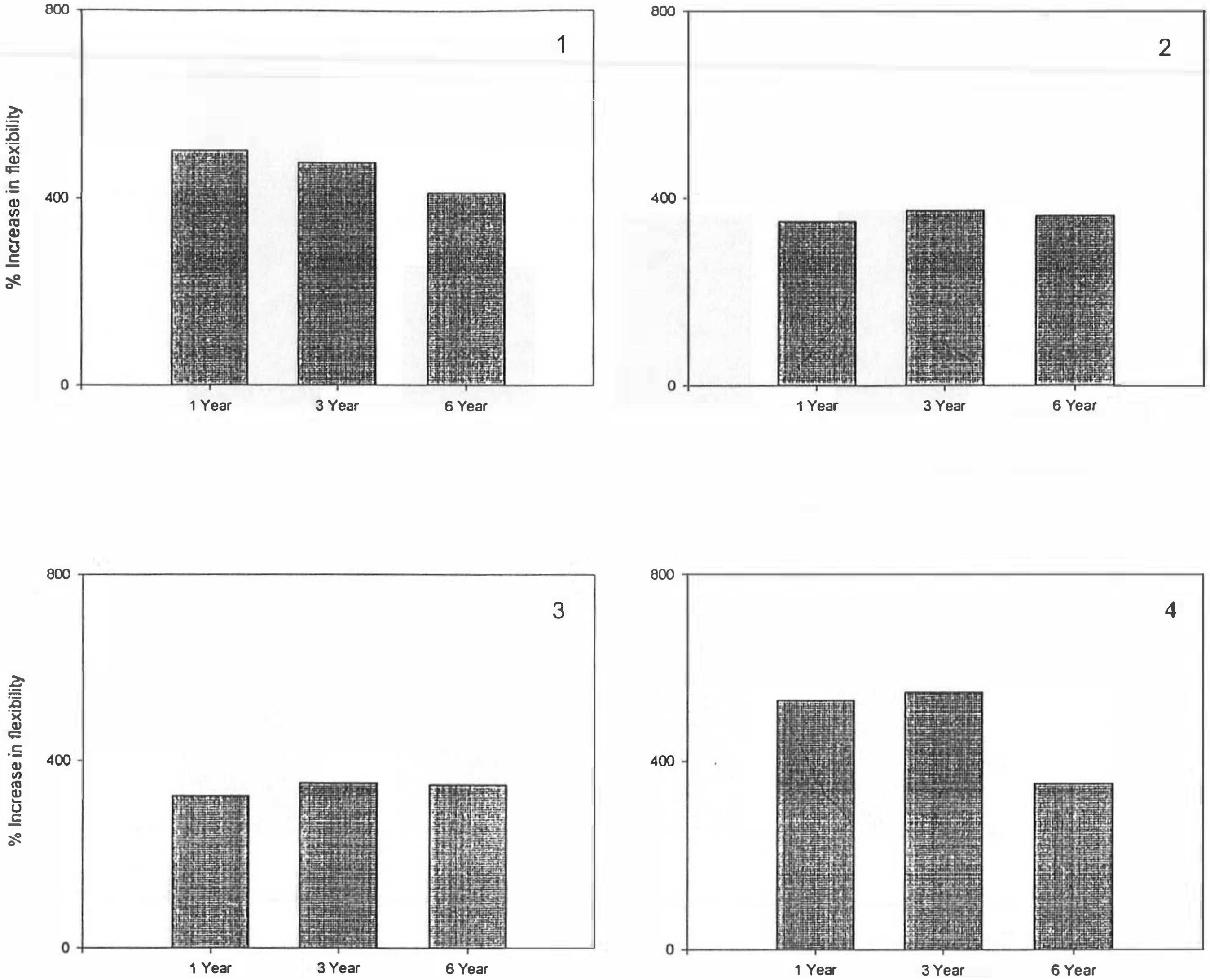
Because the present study was aimed to determine the external and internal responses of the one, three and six year old pediatric cervical spine structures, and to compare the responses with the adult human, it was necessary to create/use four specific anatomically distinct fully three-dimensional finite element models. For the adult spine, we used our anatomically accurate C4-C5-C6 finite element model (Yoganandan et al. 1996; Kumaresan et al. 1997). This model was developed using close-up computed tomography scans to obtain the bony details. Sequential close-up cryomicrotomy images from the same specimen were used to incorporate the actual soft tissue anatomy. The details of model construction, development and validation are given (Yoganandan et al. 1996; Yoganandan et al. 1997). The model included: cortical bone, cancellous bone, endplate, lamina, pedicle, spinous process and articular masses of the vertebrae. In the intervertebral discs, the nucleus pulposus, anulus fibers and ground matrix were included. Anterior longitudinal and posterior longitudinal ligaments, capsular ligaments, ligamentum flavum and interspinous ligaments were incorporated. In the facet joints, the articular cartilage, synovial fluid, synovial membrane and lateral masses were simulated. The uncovertebral joints included the synovial fluid and surrounding membrane.

The one, three and six year old pediatric cervical spine structures incorporated the appropriate developmental anatomy for these specific age groups. Briefly, these included the variations as a function of age: the spinous process cartilages, costal cartilages of the transverse processes, bilateral neural-central cartilages, rostral and caudal growth plates, lack of the uncovertebral joint anatomy together with the uncinate processes, relatively larger nucleus pulposus, differing stiffnesses for the anulus fibers embedded in the ground matrices, facet joint orientations, and various spinal ligamentous structures. Developmental processes such as bilateral neural arch and vertebral centrum ossifications were accounted for in these age-dependent pediatric models. Thus, all pediatric models included the effects of the anatomical and material changes representative of the growth and development process in the human. Our previous paper provides an overview of the developmental processes of the human cervical spine. The reader is referred to our publication for further details including element idealization and material property selection (Kumaresan et al. 1997).

All four models were exercised by fixing all degrees-of-freedom at the inferior-most vertebra. The external load was applied to the superior-most vertebra. Compression-flexion loading was applied at eccentricities of 1 cm (CF1) and 2 cm (CF2) anterior to the posterior longitudinal ligament. Similarly, the compression-extension loading was applied at eccentricities of 1 cm (CE1) and 2 cm (CE2) posterior to the posterior longitudinal ligament. In addition, pure tension load was uniformly applied on the superior surface of the finite element model. The analyses included the effect of geometric (large deformation and strain) and material nonlinearities. The external response under each load vector for each of the three pediatric models was normalized with respect to the corresponding response from the adult finite element model. In addition, the following intrinsic biomechanical variables were extracted from the finite element output under all the five loading cases. This included von Mises stresses in the anulus fibrosis in the disc and vertebral centrum/cancellous core, and the pressures in the intervertebral disc and facet joints.

RESULTS

Figures 1-5 illustrate the percentage change in the flexibility of the pediatric one, three and six year old cervical spine structures compared to the adult spine under compression-flexion (CF2, CF1), compression-extension (CE1, CE2), and tension load vectors, respectively. As can be seen, the increases in the flexibilities are a function of the type of load vector and age in the pediatric population. In general, for all loading modalities, increasing eccentricities in the load vector, whether compression-flexion or compression-extension, resulted in accentuated increases in the flexibility for all the three age groups. The internal responses expressed by the stresses in the intervertebral disc and vertebral components for the adult, and the pediatric one, three and six year old cervical structures for all five loading cases are depicted in figures 6-10. The anulus stresses, in general, were the highest for the adult spine with the highest magnitudes occurring at the most extreme compression-flexion (CF2, Figure 6) and compression-extension (CE2, Figure 9) load vectors. Furthermore, the stresses generally decreased from the adult spine to the one year old pediatric spine structures. The stresses in the vertebral components consistently had the highest magnitudes for the adult spine. In all pediatric finite element models, the facet joint pressures were always higher than the pressures in the intervertebral discs. The pressures in the disc and facet joints were respectively positive and negative for the most extreme compression-flexion (CF2, Figure 6) loading case. In contrast, the pressures were consistently positive for both compression-extension (CE1, Figure 8; CE2, Figure 9) cases and the other compression-flexion (CF1, Figure 7) loading case. For the axial tension loading case, as expected, the disc and facet joints responded with negative pressures, and the stresses in the intervertebral disc and bony components decreased with decreasing age from the adult to the pediatric populations (Figure 10).



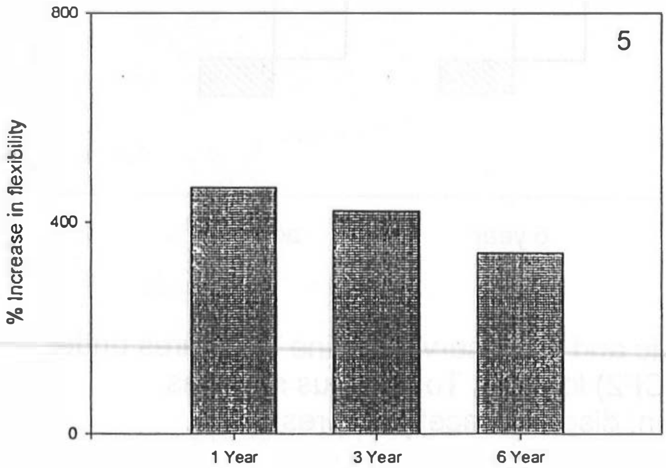


Figure 1-5: Percentage change in flexibility of the pediatric cervical spine structures:combined compression-flexion loading-CF2 (figure 1), combined compression- flexion loading-CF1 (figure 2), combined compression-extension loading-CE1(figure 3), combined compression-extension-CE2 (figure 4), and axial tension loading (figure 5).

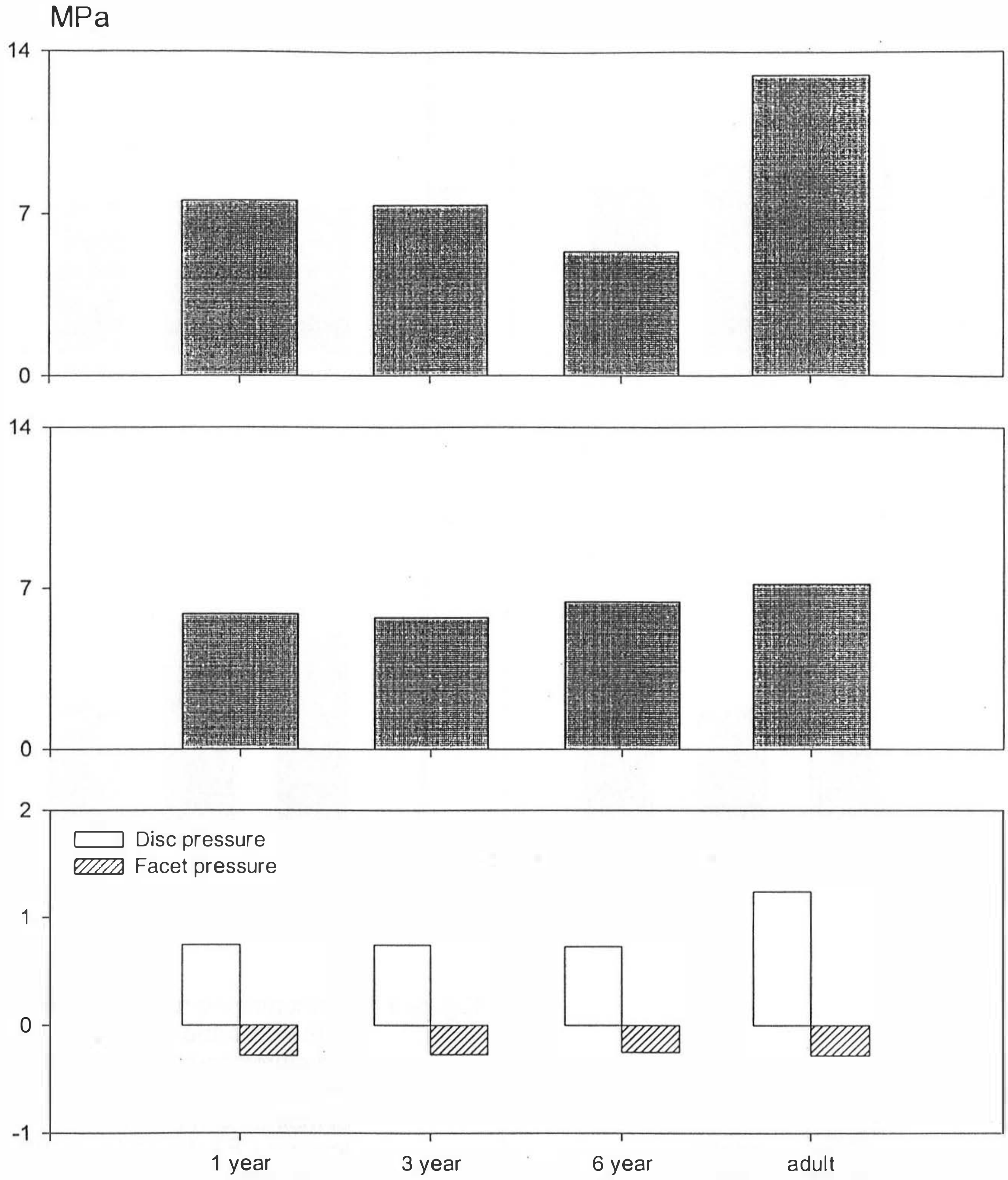


Figure 6: Internal responses of the pediatric and adult cervical spine structures under combined compression-flexion (CF2) loading. Top: anulus stresses, middle: vertebral stresses, bottom: disc and facet joint pressures.

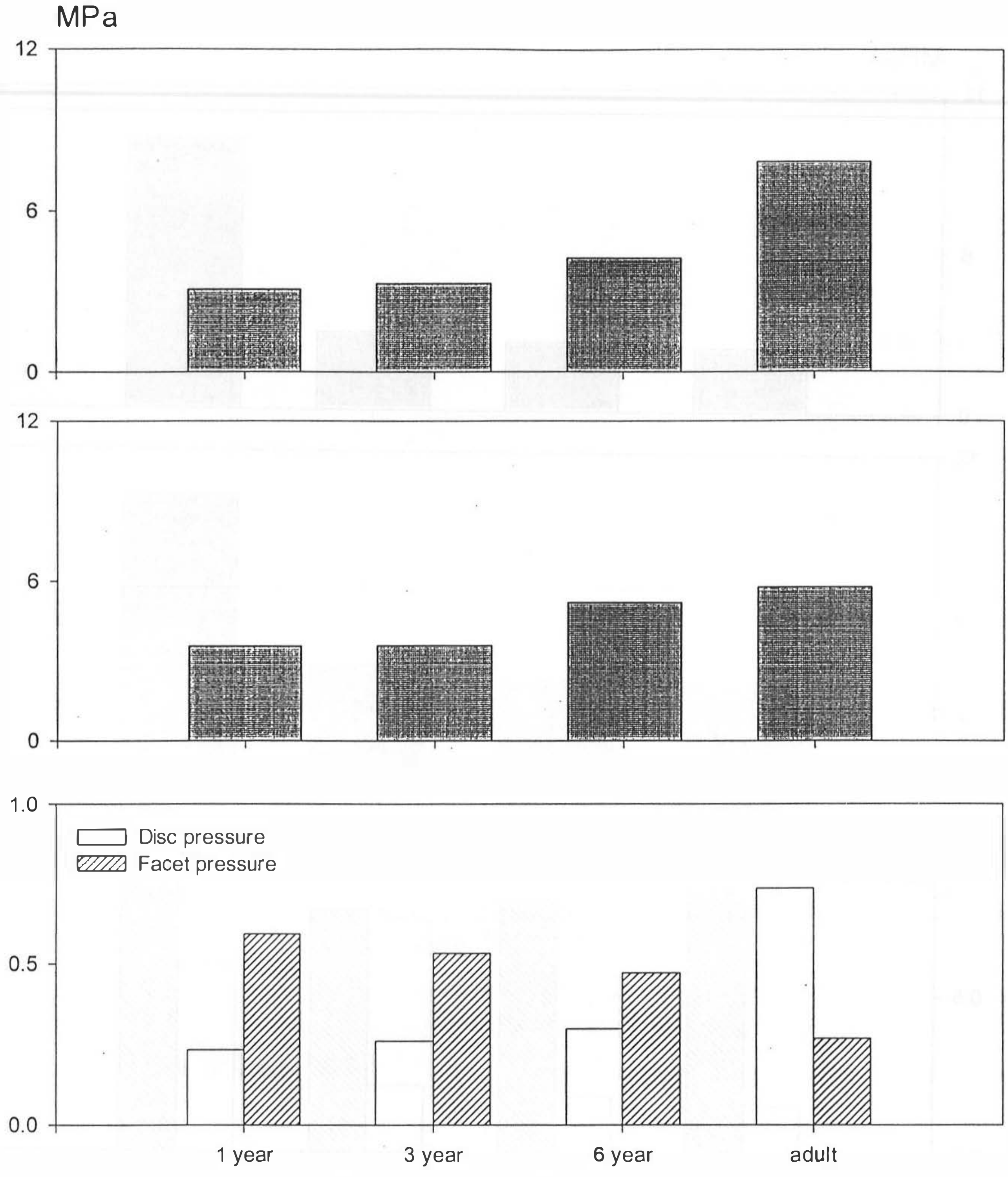


Figure 7: Internal responses of the pediatric and adult cervical spine structures under combined compression-flexion (CF1) loading. Top: anulus stresses, middle: vertebral stresses, bottom: disc and facet joint pressures.

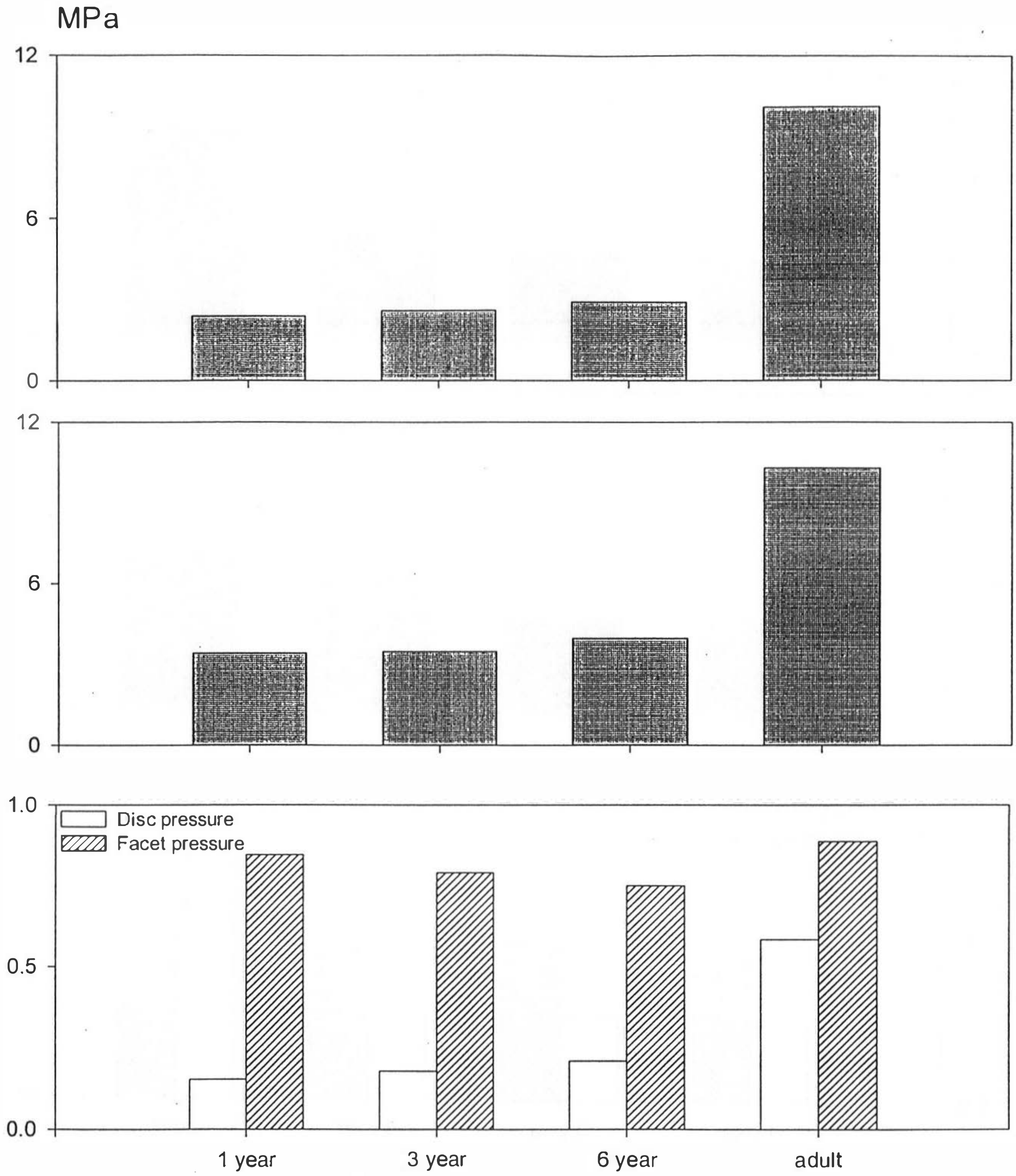


Figure 8: Internal responses of the pediatric and adult cervical spine structures under combined compression-extension (CE1) loading. Top: anulus stresses, middle: vertebral stresses, bottom: disc and facet joint pressures.

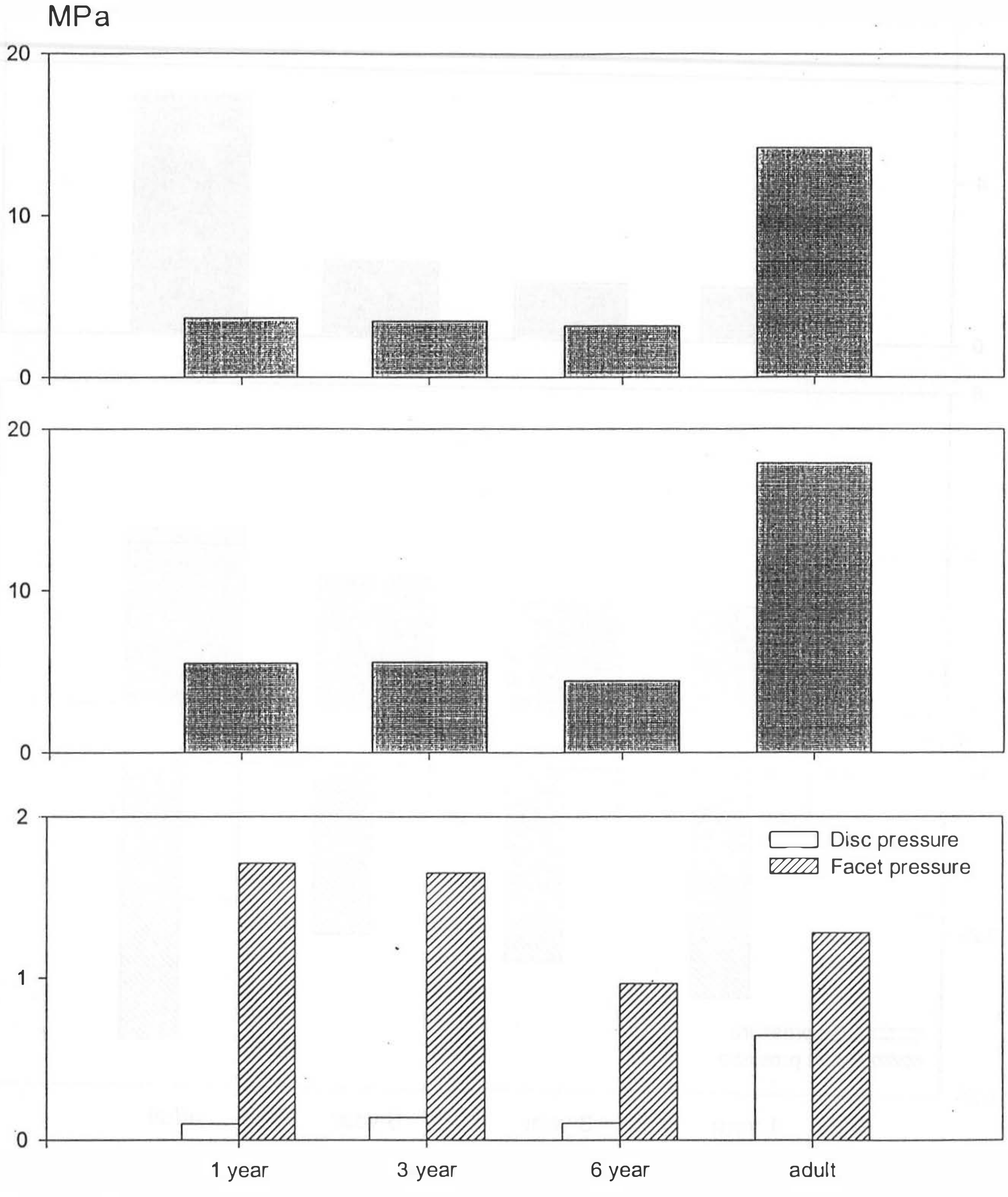


Figure 9: Internal responses of the pediatric and adult cervical spine structures under combined compression-extension (CE2) loading. Top: anulus stresses, middle: vertebral stresses, bottom: disc and facet joint pressures.

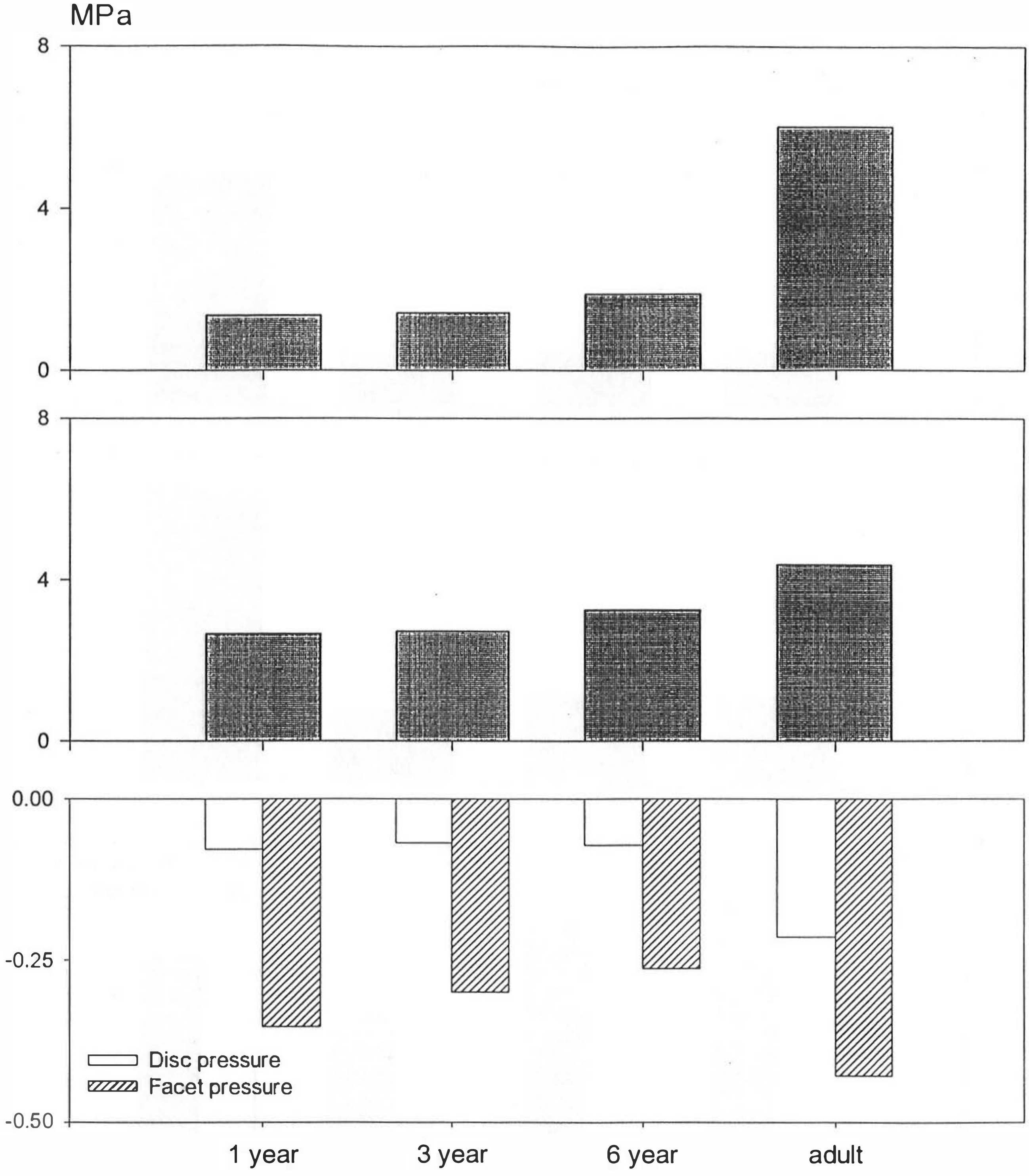


Figure 10: Internal responses of the pediatric and adult cervical spine structures under axial tension loading. Top: anulus stresses, middle: vertebral stresses, buttom: disc and facet joint pressures.

DISCUSSION

Although considerable research efforts have been advanced to develop finite element models of the human adult cervical spine structure, the application of the finite element method to solve pediatric cervical spine problems is relatively new. Using the adult finite element modeling work as a basis together with our experience in cervical spine biomechanics research, we have attempted to analyze the intervertebral biomechanics of pediatric structures using the finite element method. In our earlier publication, as a first step, we treated the pediatric cervical spine to be linearly elastic and developed a non age-specific (representative) three-dimensional finite element model (Kumaresan et al. 1998). This simplistic idealized linear model was studied under physiologic pure moment-type flexion, extension, axial torsion and lateral bending modes. Because of the simplifying assumptions used in the analysis, this finite element model approximated the intervertebral disc and facet joints as linearly elastic structures. Consequently, the incompressible nature of the nucleus pulposus, the simulation of the anular fibers embedded in a ground matrix, and the simulation of the facet joint consisting of the synovial fluid, synovial membrane and articular cartilage were excluded in this earlier research.

In order to more realistically simulate the actual spinal components such as the synovial fluid, synovial membrane, anular fibers and ground matrix, it was deemed necessary to extend this linear model into the nonlinear domain. In addition, to characterize the age-dependent behavior of the pediatric spine structures, it was further necessary to incorporate the actual developmental anatomical processes as the above "representative" single finite element model was inadequate. These modifications resulted in the development of age-specific nonlinear finite element models which were initially exercised under three simple loadings: pure compressive force, pure flexion moment and pure extension moment. The external responses were expressed in terms of the changes in the flexibilities and these results were presented at the 1997 Stapp Car Crash Conference (Kumaresan et al. 1997). The purpose of the study was to determine the differences in scaling methodologies. No internal responses (e.g., disc stresses and facet joint pressures) were analyzed in this earlier study.

The present research is an on-going extension of our previously successful mathematical modeling approach. It is well known that the principles of superposition cannot be applied for biological structures, in particular, to the human cervical spine (pediatric/adult). In other words, the response under combined compression-flexion loading cannot be determined by independently summing up the model responses obtained under a pure compressive force and a pure flexion moment. Since the human cervical spine, pediatric or adult, is eccentrically located with respect to the center of gravity of the head, pure flexion/extension moments seldom occur. Consequently, it is more physiologic to study the responses of the adult and pediatric structures under combined loading. Furthermore, since the compressive force from the head always acts on the neck, investigation of the biomechanical responses under combined compression-flexion or compression-extension loading appears to be more realistic. Because of these reasons, combined load vectors were chosen in the present study. To our knowledge, a systematic analysis of the pediatric cervical spine structures under these physiologically relevant loading conditions has not been reported in literature. In addition to these combined load vectors, as a further extension of our previous study, we subjected all the finite element models to axial tensile forces so that a comparison in the responses could be made between pure tension and pure compression. These issues formed a basis for the protocol used in the present study.

The external biomechanical responses under the tension load vector from all the cervical spine models indicated decreasing percentage changes in the flexibilities from the one year old to the six year old spine (Figure 5). The decrease in the flexibility primarily stems from the variations in the characteristics of the intervertebral discs and bony ossification centers in the developing one, three and six year old human spine. For example, the anulus fibrosis becomes stiffer as age advances enhancing the local rigidity of the disc. Likewise, the strengthening of the elastin and collagen components of the ligaments with advances in spinal growth further contribute to the decreasing flexibilities. The development of the uncinate processes together with the uncovertebral joints occurs approximately during the second decade of life. This development further reduces the flexibility of the now adolescent spine progressing toward the skeletally mature stiffness of the adult spine. All of these factors together are responsible for the extensive increase in flexibility for all the pediatric models compared to the adult. This may account for the altered trauma mechanisms in children compared to the adult. The development of neurostructures precedes the process of skeletal maturity. The neural structures are at a higher risk for trauma compared to the skeletal structures because of these developmental differences. This is manifested by the occurrence of SCIWORA (the expanded form of the acronym stated previously) only in children (Pang and Wilberger 1982)

As expected, under axial tension, the internal response determined by the pressures in the facet joint was consistently higher than the pressures in the intervertebral disc. This phenomenon was found to be true for all pediatric age groups. The negative nucleus and facet joint pressures occur because of the tensile nature of the load vector. A likely explanation for the enhanced pressures in the facet joint compared to the disc nucleus stems from the fact that the nucleus is not only surrounded by the stiffer anulus but also it is bound by the thicker growth plates. In contrast, the surrounding media for the synovial fluid of the joint capsules are relatively soft and geometrically are not as dense compared to the disc.

Under the combined loading vectors of compression-flexion and compression-extension with varying eccentricities, the disc pressures were consistently positive while the pressures in the facet capsule were positive for all cases except under extreme compression-flexion (CF2) load vectors (Figures 6-10). The progressive movement of the instantaneous axis of rotation towards the anterior regions of the intervertebral disc places the facet joints under local tension under the CF2 load vector. However, under the extreme compression-extension (CE2) load vector, the pressure inside the intervertebral disc still remains positive because of the natural placement of the nucleus posterior to the center of the disc. In fact, such movements in the location of the instantaneous axis of rotation have been identified using flexion-extension radiographs (Sherk et al. 1989). In general, similar variations in the disc and facet pressures were observed in all the three pediatric age groups. Furthermore, the pressures inside the intervertebral discs were lower in all pediatric groups compared to the facet joint pressures under compression-extension (CE1, CE2) and compression-flexion (CF1) loading cases.

Consistently, the stresses in the intervertebral discs decreased from the adult to the pediatric population for all loading cases (Figure 6-10). As expected, the stresses were the highest under extreme compression-flexion (CF2) and extreme compression-extension (CE2) loading cases; this was true for each finite element model. The variations in the intervertebral disc stresses were more magnified compared to the variations in the cancellous core stresses for the two compression-flexion loading cases. However, lesser variations were found for the compression-extension cases. The changes in these internal biomechanical responses were not uniform under each load vector for each age group of the pediatric spine. Likewise, the percentage increases in the flexibilities representing the external responses were also nonuniform. While similar changes were obtained for the less severe compression-extension and compression-flexion (CE1, CF1) loading cases, the percentage changes in the flexibilities were more pronounced for the two extreme compression-extension and compression-flexion (CE2, CF2) loading cases. The added severity of the compressive force vector in association with the flexion or the extension bending moment contributes to the accentuated increases in the flexibilities under the two extreme cases of combined loading. The percentage increases in the flexibilities under these extreme loading cases are different from the percentage increases in the flexibilities found in our earlier study under pure compression or under pure flexion/extension moments (Kumaresan et al. 1997). A plausible explanation for the greater influence of the axial component is due to the increase of cartilaginous structures which causes a degree of decoupling between bony elements. For example, the presence of bilateral neural-central and spinous cartilages in the very young essentially acts to isolate the bony load bearing structures, whereas in the adult the vertebra is more homogeneous. As the ossification centers coalesce, the facet joint orientations also angulate thus contributing to the load bearing capacity. In the present study, because the intricate developmental processes in the pediatric population were included, it was possible to delineate the external and intrinsic biomechanical characteristics of the cervical spine under various load vectors and explain the mechanisms of load transfer.

These results clearly indicate that the biomechanical responses of the pediatric cervical spines under pure bending or pure compression are considerably different from the biomechanical responses under combined loading situations. Since these changes vary as a function of the age group of the pediatric spine, it may be appropriate to consider these features in the development of anthropomorphic test devices to predict injury in crash environments. It should however, be emphasized that the results from this model are applicable to the adult and pediatric human lower cervical spine structures. In order to obtain realistic estimations of the entire human neck behavior, it will be necessary to extend this model to include additional levels of the cervical spine which are considered as future extensions. Because the present finite element models are anatomically accurate which incorporates all the load bearing structural components of the cervical spine including the hard and soft tissues with accurate representation of the various internal components, it would be a relatively easier task to extend the present model to encompass the entire pediatric human cervical spinal column.

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