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Full Length Article

Kinematical considerations related to prosthesis position and core radius on the biomechanics of the C5-C6 functional spinal unit

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ABSTRACT

Total disc arthroplasty (TDA) is one of the most preferred surgical procedure instead of fusion in the treatment of a degenerated disc. In this study, the effects of changing position and core radius of an artificial disc prosthesis, placed in the C5-C6 functional spinal unit (FSU), on the range of motions (ROM) of the FSU was investigated via the finite element method. Firstly, three-dimensional CAD models of C5 and C6 vertebrae was obtained from a computerized tomography images of healthy male neck. Finite element model of a healthy C5-C6 FSU was created containing vertebrae, intervertebral discs, tissues and facet joints considering the normal anatomical features. In modelling of TDA method, the core radius of artificial prostheses was assumed as 4 mm, 6 mm and 8 mm, and their positions were changed to anterior, posterior and lateral according to their neutral positions. Finally, the effects of core radius and position changes of the prosthesis on the biomechanical properties of the FSU were investigated by the finite element simulations. It was found that the flexion and extension ROM of the implanted FSU was increased with changing the rotation center of the prosthesis from anterior to posterior and increasing the core radius from 4 to 8 mm. However, displacing the rotation center of the prosthesis in lateral direction did not change the ROM of the model for all motion types.

1. Introductıon

Intervertebral discs are one of the significant elements of the spine, which allow bending and twisting of the vertebral column, and distribute loads to the adjacent vertebral bodies. Due to the aging, striking differences can occur in shape, structure, and composition of the discs that reduces the flexibility of the spine [1]. However, the C5-C6 functional spinal unit is the most mobile segment of the cervical spine and is therefore the most common site of disc degeneration [2]. Anterior cervical discectomy and fusion (ACDF) and total disc arthroplasty (TDA) are the surgical techniques used in the treatment of various cervical disc diseases. On the other hand, the finite element (FE) method is used commonly in the biomechanical investigations of these techniques and the artificial discs. The FE method is acknowledged as a substantial computational tool in different biomedical areas based on its ability to represent complex systems with material nonlinearities, abnormal loading, geometrical and material domains and has been widely used for biomechanical analysis of the spine [3]. The finite element models of the cervical spine were used to assess spinal biomechanics; such as material modeling effects [4], impact load effects [5] and spinal implant performances [6]. Thus, this study is aimed to investigate the effects of TDA method on the biomechanics of the C5-C6 FSU by finite element method.

The design goal of the spinal implant is to regain the healthy kinetics and kinematics of the destabilized spine with the reasons such as trauma or vertebral tumors. Therefore, in the design process of a spinal implant, the impact of the implant on global kinetics and kinematics of the spine should be considered. The interaction between the implant and the biological structure can also be explored by a FE model that mimics the physiological behavior of the spine or spinal motion segment [7,8]. Lee et al. [6] investigated the effects of an artificial disc constraint type on the biomechanics of subaxial cervical spine after TDR by the FE method. In the study, a FE model of the healthy C2-C7 spine segment was developed and the model was validated with previous studies. Then, the effects of core type on the biomechanics of the cervical spine segment was investigated with two different types of an artificial disc (fixed and moving cores) implanted at the C5-C6 level. They concluded that, the range of motion (ROM), ligament stress and facet joint forces at the TDR segment increased after implantation of an artificial disc prosthesis and the increment ratio

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of the parameters is found higher in mobile-core model than fixed-core model. Li et al. [9] investigated the effects of ball-and-socket type prosthesis geometry on the biomechanics of C3-C7 cervical segment after hybrid surgery (HS). In their study, they examined two different hybrid surgery (HS) methods applied on C4-C6 cervical segment by the FE method. In the first method, while applying anterior cervical discectomy and fusion (ACDF) to the C4-C5 level, the anterior cervical disc replacement (ACDR) method was applied to the C5-C6 vertebra level. The performances of three different prosthesis designs were examined in the ACDR applied level. In the second HS method, they changed the surgical methods applied to the segments with the other. At the end of the study, they proposed the first investigated HS method as an optimal treatment for decelerating the adjacent segment degeneration at C4–C6 level [9]. The influences of positioning and design parameters of the implant on the biomechanics of the C5-C6 spine unit was investigated by Galbusera et al. [10]. In the study, axial, antero-posterior and lateral positions of the implant center of rotation were considered as probabilistic variables. At the end of the study, an appropriate positioning of the implant, especially in the antero-posterior direction, was emphasized as a critical importance during the surgery. Wheeldon et al. [11] developed a finite element model of C4-C7 cervical spine segment and validated the model with the experimental data taken from the flexion-extension, lateral bending, and axial rotation loading tests of nondegenerated cervical spines. In the study, the mean and standard deviation corridors of an experimental test was suggested as an ideal tool for the validation process of finite element models.

The aim of this study is the determination of the biomechanics of a cervical FSU implanted with a ball and socket type artificial disc prosthesis in comparison with the biomechanics of the healthy FSU. For this purpose, 3D FE model of the healthy C5-C6 FSU was developed based on the computerized tomography (CT) images of healthy human neck. The biomechanics of healthy and implanted models were compared, and the simulation results were discussed in terms of clinical significance of prosthesis rotation center location and core radius on the biomechanics of the implanted FSU.

2. Material and methods

In this study, the effects of changing the prosthesis position and core radius on the biomechanical behavior of a C5-C6 FSU was investigated by a FE model implanted with a ball and socket type disc prosthesis. Firstly, three-dimensional CAD models of C5 and C6 vertebrae was obtained from a computerized tomography (CT) images of 45-years old healthy male neck by using 3D Slicer software (https://www.slicer.org/). A 3D finite element (FE) model of the healthy C5-C6 FSU was developed with the ligamentous tissue including the bony sections (cortical, cancellous, and posterior), vertebral disc, cartilaginous end plates, facet joints, and five primary ligaments. The bony sections of the vertebrae and the intact intervertebral disc were meshed with hexahedral eight node solid elements. Meshing of the bony sections was made with Bolt Software (Csimsoft, Utah, USA). A total of 51,734 solid elements were used for meshing bony sections and intervertebral disc. The intervertebral disc was divided into two parts as the central nucleus surrounded by an annulus, and the volume of the nucleus was defined as about 40% of the total disc volume, as given in Ref. [2]. Cartilaginous endplates having an average thickness of 0.6 mm were modeled to enclose the superior and inferior faces of the intervertebral disc. The gelatinous property of the nucleus was defined via a nearly incompressible elastic material.

The mechanical behavior of the annulus has been investigated by two sort of models named as structural finite element models and fiber-reinforced strain energy models. In the first model the anisotropy is defined by using truss or cable elements, in the second model anisotropy is defined by invariant mathematics and directional tensors [12]. In this study, Holzapfel-Gasser-Ogden (HGO) material model, which is a fiber-reinforced strain energy model and developed for the modeling of anisotropic hyperplastic materials, is used to describe the mechanical properties of annulus. This material model is extensively utilized in the FE models for simulating the elasticities of fiber reinforced biological materials [13] and splits the Helmholtz free strain energy function (Ψ) into different components corresponding to the ground substance, the fibers, and the material compressibility. It is described as below [7].

$$\Psi(C, A_1, A_2) = \bar{\Psi}_{gs}\left(\bar{C}\right) + U(J) + \bar{\Psi}_f\left(\bar{C}, A_1, A_2\right)$$

$$= C_{10}\left(\bar{I}_1 - 3\right) + \frac{1}{D}(J-1)^2 + \frac{K_1}{2K_2}\sum_{i=4,6}\left\{exp\left[K_2(\bar{I}_i - 1)^2 - 1\right]\right\}$$

where, D and C_{10} are the compressibility and stiffness related parameters of the ground substance, respectively, $K1 > 0$ and $K2 > 0$ are material parameters. $\bar{I}_1$ is the first deviatoric strain invariant of modified Cauchy–Green tensor, J is determinant of the deformation gradient, and the anisotropic behavior of the fibers are characterized by $\bar{I}_i$ (i = 4,6) invariants defined as square of fiber stretches in the orientation directions. The tensile properties of the annulus greatly depend on the position within the intervertebral disc. The tensile modulus and failure stress of the anterior region of the annulus is bigger than the posterolateral region and also, the inner regions of the annulus have greater failure strains and lower tensile modulus and failure stress than the outer regions [14]. Therefore, the annulus is divided into four parts as anterior inner (AI), anterior outer (AO), posterior inner (PI) and posterior outer (PO). The material properties for the sections of annulus is taken from the study of Rao [15] and given in the Table 1.

To perform analytical researches on the biomechanical behavior of spinal segments, it is required to know the mechanical properties of the ligaments and also the properties of the other soft tissues including the intervertebral discs, cartilages as well as the realistic geometry of the bony structures. Significant changes in the mechanical properties of each spinal ligament were stated in the literature. Five primary cervical ligaments consisting of anterior longitudinal, posterior longitudinal, capsular, ligamentum flavum, and interspinous ligaments were modeled with spring elements acting only in tension. The material properties of the ligaments were defined by force-displacement curves generated from the constants given by the Mattuci et al. [16] for the quasi-static loading case of the cervical ligaments. The attachment points of ligaments to the vertebral bones were defined depending on the human anatomy [2]. The used material properties for defining the bony structures, ligaments and intervertebral discs were given in the Table 2.

In the modeling of facet joints the cartilage layer on the joint surfaces represented by an exponential contact pressure-overclosure relationship as given in Ref. [18]. In this relationship the contact pressure on the facet joints increases exponentially based

Table 1
Material constants of the strain energy function for the different annulus regions.

Parameter	AO	AI	PO	PI
C_{10} (MPa)	0.0931	0.0914	0.0093	0.5383
K_1 (MPa)	253.18	259.68	759.46	485.82
K_2	569.69	905.17	331.76	500.45
D $(MPa)^{-1}$	2.2656	2.5946	1.0321	489.18

Table 2
The material properties used in the C5–C6 FE model.

Component	Constitutive model	Material properties	Element Type	Reference
Cortical bone	Isotropic elastic	E = 12000 MPa, ν = 0.3	Hexahedral	[11]
Cancellous bone	Isotropic elastic	E = 100 MPa, ν = 0.2	Hexahedral	[11]
Posterior bone	Isotropic elastic	E = 3500 MPa, ν = 0.25	Hexahedral	[11]
Cartilaginous endplate	Isotropic elastic	E = 23.8 MPa ν = 0.4	Hexahedral	[17]
Nucleus Pulposus	Isotropic elastic	E = 1 MPa, ν = 0.49	Hexahedral	[17]
Annulus fibrosus	Anisotropic Hyperelastic	Table-1	Hexahedral	[15]
Ligaments		Nonlinear	Connector	[16]
Facet Cartilage	Non-linear soft contact			[18]
Artificial discs		Kinematical connection	Connector	[19]

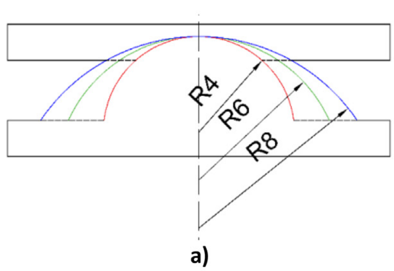
on the decrease in the distance among the two facet surfaces. When the joint surfaces contact each other, the maximum allowable contact pressure on the surfaces was defined equal to the elasticity modulus of the cortical bone.

In the modeling of the TDA method, the intervertebral disc and the anterior longitudinal ligament were totally removed from the FE model of the healthy FSU. Due to advantages of kinematic coupling method in reducing modification and calculation time of the FE model, and changing the spinal kinematics slightly [19], the prosthesis was modeled with connector elements instead of conventional volumetric elements. Therefore, the ball and socket type artificial disc having a fixed rotation center is modeled by a connector (join and rotation type) element. The inferior end plate surface of the C5 vertebra and the superior end plate surface of the C6 vertebra were kinematically connected at a reference point representing the rotation center of the prosthesis. With the kinematic coupling, the lift-off or separation phenomenon arising from loss of contact between the prosthesis upper plate and the core [20], is not permitted in the prosthesis. The center of the healthy disc was assumed as the neutral implanting position of the prosthesis. The positions of the disc are changed in anterior, posterior and lateral directions according to the neutral position. The rotation center locations of the prosthesis were changed randomly in turn from the neutral position as 2 mm for the anterior, 1.5 mm for the posterior and 1 mm for right-left laterals. The core radius of the disc was assumed as 4 mm, 6 mm, and 8 mm. Hence, to represent the radius change of the prosthesis in the implanted positions, the coupling point of the connector element was shifted axially as described in Fig. 1. The generated FE model of the healthy FSU is given in the Fig. 2.

The FE simulations were performed by ABAQUS/Standard 6.11 software (Simulia, Providence, RI, USA). To perform FE simulations of flexion-extension, axial rotation and lateral bending movements the loading moment of 2.0 Nm was applied to the model. In the simulations, the moment loads were conducted to the model through a refence point constrained to the superior end plate of C5 vertebra and the nodes on the inferior end plate surface of C6 were constrained during the analyses. The healthy C5-C6 FSU model was validated using the biomechanical data given by Wheeldon et al. [11,21] as an experimental corridor of cervical flexion-extension, lateral bending and axial rotation simulations. In the second stage of the study, the effects of changing the rotation center position and core radius of an artificial disc prosthesis on the biomechanics of C5-C6 functional spinal unit were investigated by the FE method. The core radius of the discs is changed from 4 mm to 8 mm in the FE models. Finally, a comparison was made between the simulation results of healthy and implanted functional spinal units to determine the effects of prosthesis location and core radius on the biomechanical properties of the C5-C6 FSU. In the modeling of a prosthesis core radius change, the rotation center of the prosthesis was shifted axially as described in Fig. 1. Subsequently, these rotation centers were changed randomly from the neutral positions to 2 mm in the anterior, 1.5 mm in the posterior, and 1 mm in the right and left laterals as shown in Fig. 1.

3. 3. Results and discussion

A finite element model of the healthy C5-C6 FSU was constructed, and simulations were carried out to study the moment



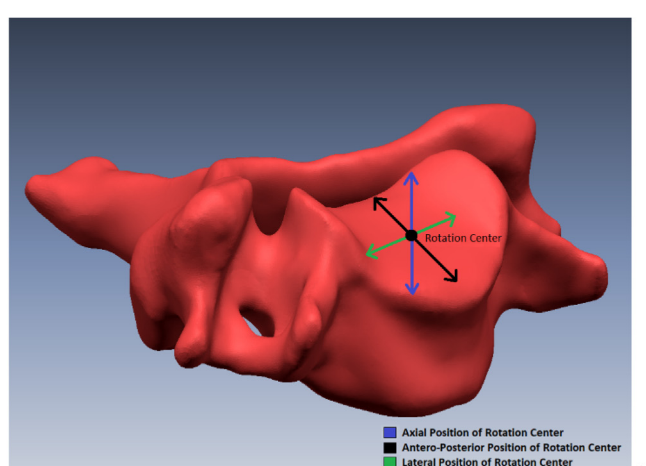


Fig. 1. (a) Schematic representation of rotation centers of the prosthesis socket section, (b) The changing of rotation centers in the axial, antero-posterior and lateral directions.

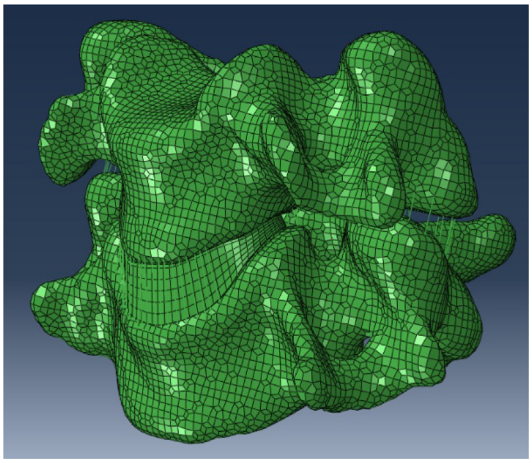


Fig. 2. Finite element model of the C5-C6 FSU.

rotation behavior of the model under the flexion-extension, lateral bending and axial rotation movements. The comparison curves, obtained from in vitro experimental studies using similar loading and boundary conditions, were used to validate finite element simulations.

The findings from the experimental study of Wheeldon et al. [21] is used to validate the simulations of flexion-extension motions and the plotted comparison graph for the moment-rotation is given in the Fig. 3. On the consideration of moment rotation curves for the investigated motions, the flexion motion exhibited a lower stiffness according to the extension motion in agreement with the compared study. In the simulations of the extension motion, closer results are obtained to the average moment-rotation curve in the compared study. In the case of the flexion motion, the results obtained from the FE simulation showed a little more stiffness than the compared study, but the moment rotation values obtained from the simulation remained within the standard deviation limits of the compared study. As a result, the flexion and extension simulations of healthy C5-C6 FSU showed that the obtained moment-rotation characteristics are consistent with the study of Wheeldon et al. [21]

In the validation process of the axial rotation and lateral bending simulations, a comparison was made with an experimental corridor given by Wheeldon et al [11]. The moment-rotation behavior of the healthy C5-C6 model from simulations of the axial rotation and lateral bending is given in Fig. 4. As seen from this figure, the

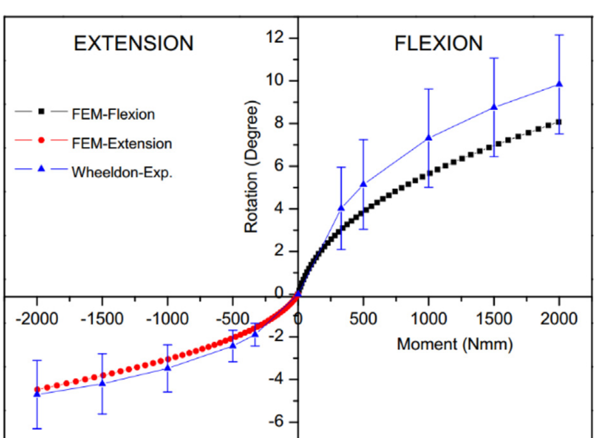


Fig. 3. The moment-rotation behavior of the C5-C6 intact segment for the flexion-extension motion.

predicted ROM of the healthy C5-C6 model for the lateral bending and axial rotation simulations were found within the experimental corridor boundaries of referred study [11]. On the other hand, the moment-rotation behavior predicted by the FE model in the lateral bending simulation exhibited a stiffer behavior than the compared average moment-rotation curve, while the results in the axial rotation simulation were obtained closer to the average curve. It was understood that the full [22] or partial [10] segment of a cervical spine were investigated in FE method based studies. The results are given as the load-displacement curves obtained from flexion-extension, lateral bending and axial rotation motions. In these studies, the mechanical properties defined for bones and other tissues in the FE model are generally taken from previous studies in the literature, and validation study was performed based on this material data. On the other hand, in some studies, the basic material properties taken from the previous studies were assigned to the FE model [23], but the initially assigned material properties were calibrated based on the reported ranges in the literature by using the motion data obtained from in-house experimental study [24]. In our study, the mechanical properties assigned to the bony and ligamentous structures were taken from the literature, and the FE model was not calibrated in accordance with to any experiment or data. In the studies of Wheeldon [11,21], the load-displacement curves of flexion-extension, lateral bending and axial rotation motions are given as statistical corridors, and these standard deviation corridors were used as a basis for verifying our FE model. It is also well known that the human cervical spine shows nonlinear force-displacement response, with increasing stiffness at higher loads. Generating a realistic FE model that can effectively mimics this overall nonlinear behavior of the cervical spine is an important issue [25]. The developed FE model estimated the nonlinear behavior of the C5-C6 FSU for flexion- extension, lateral bending and axial rotation motions in our study. Additionally, the extension motion exhibited stiffer behavior than the flexion motion, and so lower ROM's obtained for extension motion as in the compared study. This was attributed to facet loading, facet joint orientations, and initial position, as can be seen Ref. [26]. Moreover, the load-displacement response of the FE model was within the standard deviation corridors of the referenced study [11,21], in all investigated motion cases.

In the second step of this study, the effects of changing the rotation center position and core radius of an artificial disc prosthesis on the biomechanics of C5-C6 functional spinal unit were investigated by the FE method. Initially, the simulations of flexion-extension, axial rotation and lateral bending movements were performed for the implanted models with neutral prosthetic positions. The axial rotation and lateral bending motions were evaluated by taking average of ROM values obtained in right and left directions. The ROM comparison of the implanted and healthy models are given in Fig. 5.

Fig. 5 shows that in the neutral prosthesis position, the range of motions (ROM) of the implanted models are lower than the healthy model in flexion motion. On the other hand, the ROM of the C5-C6 FSU is reduced by decreasing the core radius of the prosthesis during the flexion. In the extension motion, it is observed that the ROM of the prosthesis having an 8 mm core radius is higher than the healthy model and the prosthesis having a 6 mm core radius is roughly the same as the healthy model. When the influence of the core radius on the ROM of the implanted model was examined, it is seen that the ROM of the implanted model is reduced with decreasing core radius of the prosthesis for both the flexion and extension motions. It is apparent in the axial rotation and lateral bending simulations that the ROM of the implanted models are higher than healthy model, and also the radius variation has negligible effect on the ROM. In a study investigating the effects of TDR method on the biomechanics of the implanted segment, the

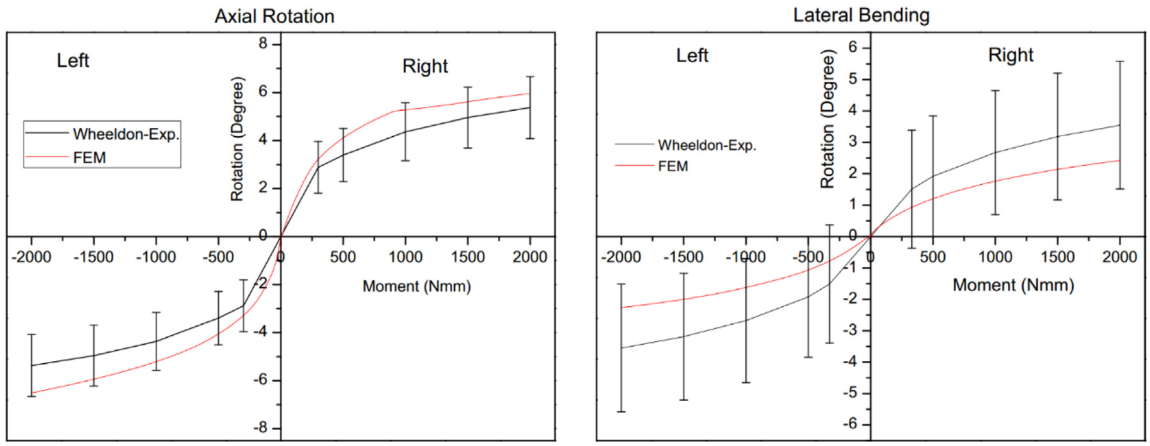


Fig. 4. The moment-rotation behavior of the C5-C6 intact segment for the right and left axial rotation, and lateral bending motions.

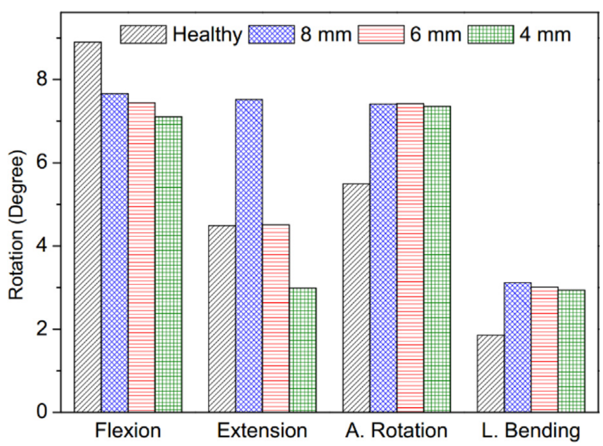


Fig. 5. ROM comparison of implanted models with the intact model for the neutral prosthesis position.

implantation of a ball and socket type artificial disc was found to provide a segmental mobility as well as the healthy spine [27]. However, in most of these studies, the existing commercial artificial prosthesis were modeled as volumetric elements and their effects on the biomechanics of the implanted segment were examined. In our study, the investigated prosthesis was not based on any commercial artificial disc. Only the kinematic connection of the prosthesis was considered, and the effects of a ball and socket type artificial disc with fixed rotation center on the mobility of C5-C6 FSU were investigated. In the neutral prosthesis position, except for the flexion motion, the higher ROM values were obtained for the other investigated motions. It is also reported that a lift-off phenomenon occurs during the motion of the model implanted with the ball and socket type artificial disc [6]. In the study, the modeled prosthesis has a fixed center of rotation, and did not allow the lift-off phenomenon due to its kinematic connection. The implanted model is thought to yield a lower flexion ROM than the healthy model due to the kinematic constraint did not allow the prosthesis to lift-off.

Finally, the rotation centers of the prosthesis were displaced from the neutral positions to 2 mm in the anterior, 1.5 mm in the posterior and 1 mm in the laterals. The range of motions obtained from the flexion and extension simulations, performed by changing the prosthesis core radius and the rotation centers from neutral to the anterior, posterior and laterals, is given in Figs. 6–7. In the flexion motion, when the rotation center of a prosthesis is changed from anterior to posterior, the ROM of the implanted model increased at all core radius values. However, when the core radius of the prosthesis was increased at the implanted position, the flexion ROM of the model is also increased. In the case of changing the prosthesis rotation center along the lateral line, it is observed that flexion ROM of the implanted model is not affected by the lateral position change of the prosthesis, and only effected by the core radius of the prosthesis. As in the flexion motion, the ROM of the extension motion increased when the center of rotation of the prosthesis was changed from anterior to posterior position. In addition, the ROM of the implanted model increased with increasing the core radius of the prosthesis. While the rotation center of prosthesis is displaced laterally, a significant change was not observed in the extension ROM. However, when the effect of prosthesis core radius on the flexion and extension ROM is examined, it is observed that the core radius of the prosthesis has more effective on extension motion.

The effects of rotation center and core radius change of the prosthesis on the ROM of the FSU for lateral bending and axial rotation motions are given in Figs. 8–9. A noticeable variance was not occurred in the ROM of the lateral bending motion, when the core radius of the prosthesis was kept constant and the rotational center positions was changed from anterior to posterior and right to left lateral. In the event of increasing the core radius of the prosthesis at the implanted position, the ROM of the lateral bending motion increased slightly.

In the axial rotation, changing the prosthesis position was not altered the ROM of the model significantly as in the lateral bending simulation. When the prosthesis rotation center was shifted from anterior to posterior position, it is seen that the using of prosthesis with different core radius was not change the axial rotation ROM. Whereas, the core radius of the prosthesis has a little effect on the ROM of the implanted model when the prosthesis rotation center was changed from right to left.

In the study, the rotation center of the prosthesis was shifted in axial direction for simulating the core radius change of the implant. In the studies of Galbusera et al. [10] and Rohlmann et al. [28] the rotation center position of a prosthesis in the axial and the antero-posterior directions were stated as the most effective parameters on the intervertebral rotation. From the FE simulations of the implanted C5-C6 model, in flexion-extension motion, we observed that the flexibility of the model influenced with the core radius and antero-posterior position of the prosthesis. However, the mobility of the implanted model remained approximately same in lateral bending and axial rotation motions by changing these the two parameters of the implant. Galbusera et al. [10] stated that, during

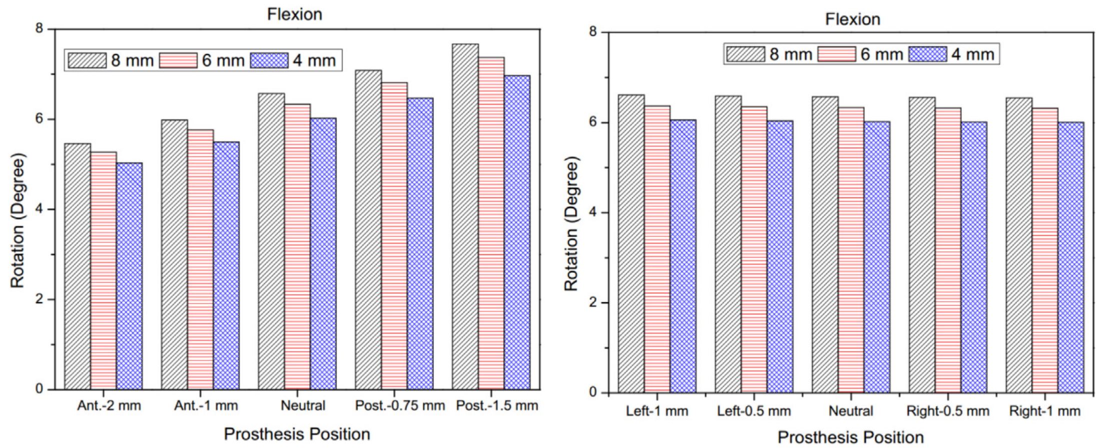


Fig. 6. The effects of prosthesis core radius and rotation center positions on the flexion ROM of the implanted models.

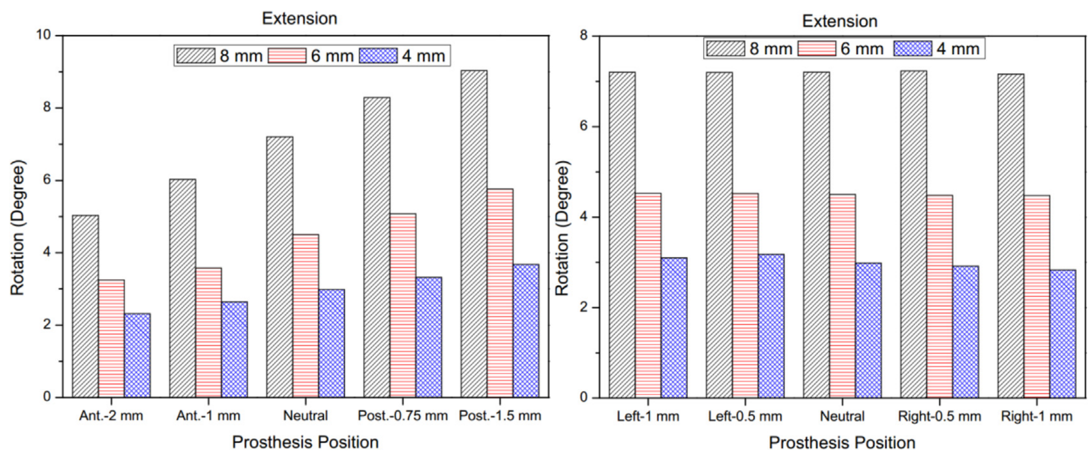


Fig. 7. The effects of prosthesis core radius and rotation center positions on the extension ROM of the implanted models.

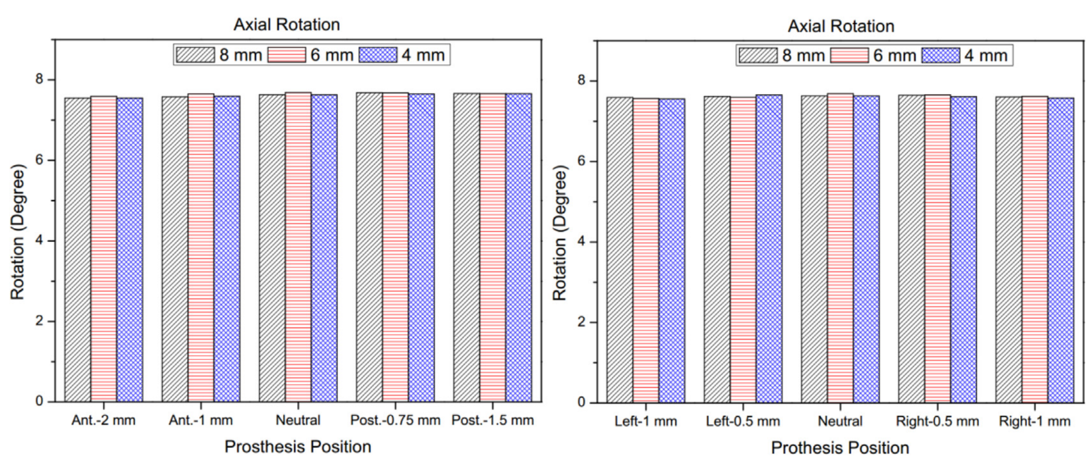


Fig. 8. The effects of prosthesis core radius and rotation center positions on the axial rotation ROM of the implanted models.

the surgery, correct positioning of the prosthesis in the antero-posterior direction may be more effective than the positioning in the lateral line in terms of affecting the biomechanics of the implanted spine. Similarly, the ROM of the implanted model was not changed significantly with the lateral position of rotation center in our study.

In the study, modeling of the artificial disc with kinematic coupling method instead of volumetric elements restricts the acquisition of stresses on the disc. Lee et al [27]., examined performances of two commercial artificial disk having different kinematic connection types (constrained and unconstrained) in the full cervical model by FE method. They stated that the constrained type disc exhibited lower ROM and higher contact stresses on the PE core surface than the unconstrained disc. In the contact mechanics of ball and socket type prostheses, the equations given about the contact pressure indicate that the contact pressure will decreases as

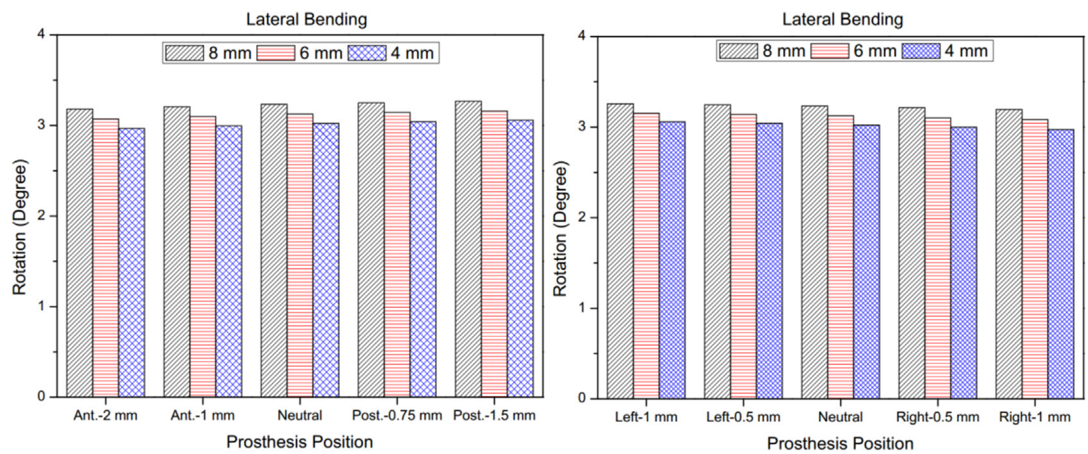


Fig. 9. The effects of prosthesis core radius and rotation center positions on the lateral bending ROM of the implanted models.

the radius of the contacting surfaces increases. Low contact pressure will reduce the core wearing of an artificial disc and so reduce the wear debris related complications [29]. Therefore, increasing the core radius at the same prosthesis height would be better in terms of design because of providing low contact pressure and higher flexion extension ROM.

4. Conclusions

In this study, the effects of changing the rotation center position and core radius of the prosthesis on the biomechanical behavior of the C5-C6 FSU was investigated by FE method. The flexion-extension, lateral bending and axial rotation motions of C5-C6 FSU were investigated for the healthy and implanted models. The simulation results of healthy model were confirmed by comparing with the biomechanical data given in the literature. The obtained findings are summarized below.

- In the flexion motion, the implanted models exhibited more stiffer behavior than the healthy model. Therefore, lower flexion ROM's were obtained in all core radius values than healthy model for neutral prosthesis position. In addition, the greater lateral bending and axial rotation ROM were obtained in the implanted models than the healthy model.
- The flexion and extension ROM's of the implanted FSU was increased with changing the rotation center of the prosthesis from anterior to posterior, and when the core radius is increased from 4 to 8 mm.
- The flexion and extension ROM's of the model did not change with displacing the prosthesis rotation center laterally.
- In the lateral bending and axial rotation motions, any change was not occurred in the ROM of the FSU when the rotation center position was changed from anterior to posterior and also right to left lateral.
- Overall results shown that correct positioning of the prosthesis in the antero-posterior direction is more effective than the positioning in the lateral line in terms of biomechanics of the implanted spine during the surgery.

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