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Original paper

# Numerical model of the human cervical spinal cord – the development and validation

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The influence of mechanical load on the extent of nervous tissue damage in the spinal cord at the time of trauma is presently incontestable. Although numerical modelling cannot fully replace physical testing, it seems to be the perfect complement to experiments in terms of the analysis of such a complex phenomenon as traumatic spinal cord injury. Previous numerical models of the human cervical spinal cord have been limited by several factors: two-dimensional modelling, spinal cord geometry simplification and incomplete reflection of specific anatomical and biomechanical relations of the objects being modelled.

The objective of this study was to develop and validate an accurate and universal numerical Finite Element Method (FEM) model of the human cervical spinal cord. Our survey focuses mainly on geometric, constraint and material aspects. Experimental validation was carried out based on a controlled compression of the porcine spinal cord specimens. Each stage of compression was simulated using the FEM model of the compressed segment. Our 3D numerical simulation results compared with experimental results show a good agreement.

It is possible to use the developed numerical model of the human cervical spinal cord in the biomechanical analysis of the spinal cord injury phenomenon. However, further clinical evaluation is clearly justified.

*Key words*: *3D modelling, cervical spinal cord injury, experimental validation, Finite Element Method (FEM)*

## 1. Introduction

The concept of modern therapy for traumatic spinal cord injury (SCI) is based on an idea of suppressing or eliminating the effects of secondary SCI. This phenomenon is principally based on an inflammatory reaction as well as on toxic effects of free radicals and nitric oxide, leading to a progressive degeneration of the previously undamaged nervous tissue of the spinal cord and an increase of neurological deficits [1]. The dysfunction of the blood–spinal cord barrier is considered to be the factor initiating the cascade of secondary SCI [2]. In 2007, MAIKOS and SHREIBER showed that the extent of the damaged blood–spinal cord barrier (BSB) was strongly correlated with the strength of the mechanical impulse damaging the spinal cord [3]. A present state of knowledge allows one to think that numerical modelling of the processes accompanying SCI can provide insight into the spatial distribution of the parameters of the state of the strain generated by the mechanical load at the time of the spinal cord injury. By providing information concerning the expected size of the damaged BSB area and the secondary spinal cord damage it could be possible to forecast the clinical course of the disease in patients suffering from SCI [4].

Attempts to carry out numerical modelling of the biomechanics of the spinal cord were made with the help of the Finite Elements Method (FEM) by numerous researchers [1], [5]–[9]. These authors were the pioneers in the modern three-dimensional (3D) FEM modelling of the spinal cord injury. However, they created simplified models and used them for hypothe-

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sized, idealized injuries to the spinal cord, which made the interpretation of the results relatively difficult. The development of a technologically advanced and universal FEM model of the spinal cord can provide a variety of valuable information. It can also be a starting point for studies on modern diagnostic and therapeutic algorithms for the successful treatment of patients with SCI.

The aim of our study was to develop and experimentally validate an accurate, universal and reproducible 3D numerical model of the human cervical spinal cord which could be used in the clinical analysis of the SCI phenomenon, based on the FEM.

# 2. Materials and methods

A three-dimensional numerical model of the human cervical segment of the spinal cord was created using the ANSYS Multiphysics, versions 11.0 and 12.1, software (ANSYS, Inc. USA).

## 2.1. Geometry of the model

The diagram of the procedure for creating a 3D model of dural sac and spinal cord is presented in figure 1. In order to avoid the disturbances of the spinal cord geometry caused by trauma (oedema, laceration, etc.), the algorithm was based on the measurements above and below the place of injury. Average axial and transverse dimensions of the spinal cord measured above and below the place of injury in a given individual were used to adjust the geometry of the model to the patient's anatomy.

The model of the spinal cord comprised two components – the model of the white matter and the model of the grey matter. Due to insufficient resolution of the MRI images of the spinal cord, exact measurements of the grey matter could not be performed.

During the analysis of the measurements of fresh axial microscopic sections from human corpses (preserved, sectioned serially and documented photographically in the Visible Human Project (VHP)) – the relationships between the sagittal and transverse diameters of the section of the spinal cord and the grey matter were observed (figure 2) [10].

The geometry of the denticulate ligament was designed based on an available anatomical data [11]–[13]. A single ligament was triangular in shape, its wide base was attached to the lateral surface of the spinal cord, and its apex directed towards the dural sac. Seven ligaments were placed symmetrically along the side surfaces of the spinal cord. The height of each ligament corresponded to the height of the spinal cord segment, whereas the place where two consecutive ligaments met corresponded to the border between two consecutive segments.

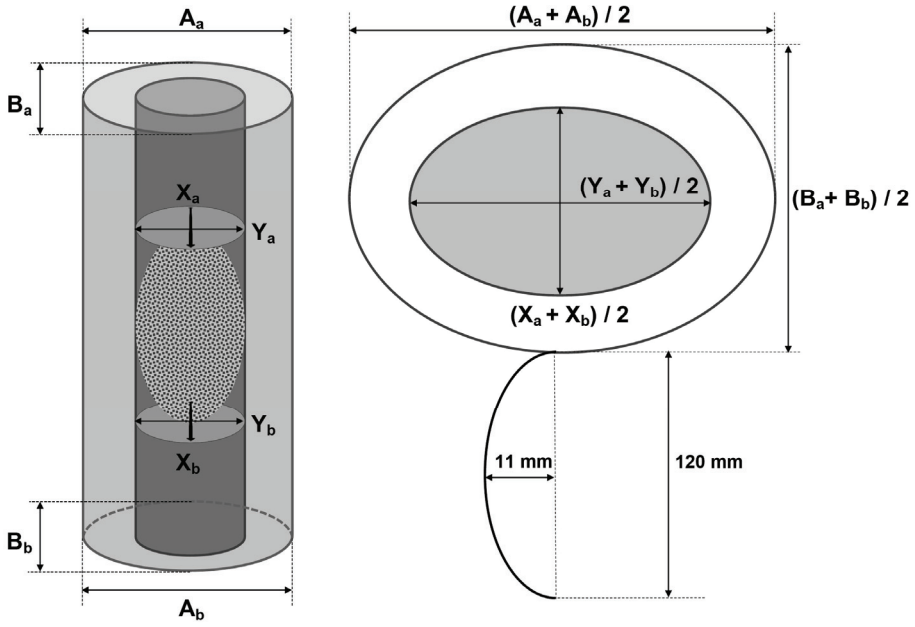


Fig. 1. Procedure for creating a 3D model of dural sac and spinal cord.
Left side: measurements taken from MRI of the particular patient above and below the site of injury (dotted area).
Right side: dural sac and spinal cord modelled as tubes of an elliptic shape in an axial plane.
The tubes were stretched on an arch of a 120 mm length and 11 mm radius [29]

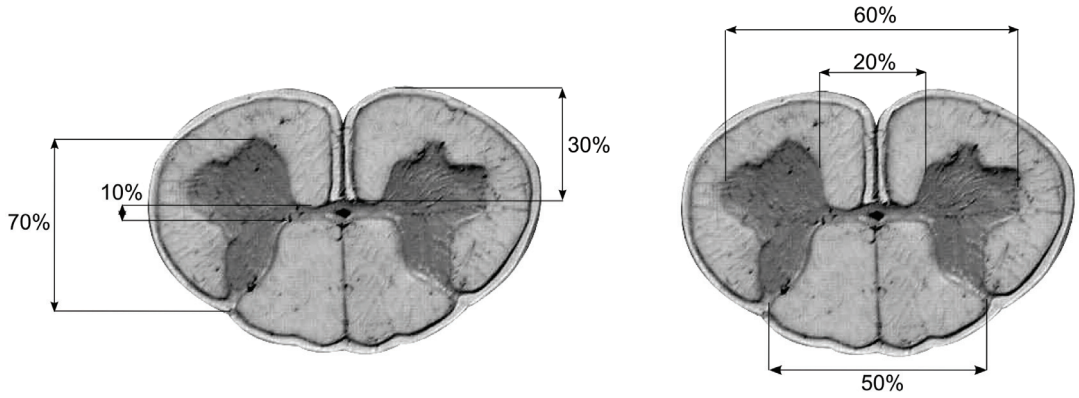


Fig. 2. Percentage geometrical relationships between grey and white matter related to:
left side – sagittal diameter of the spinal cord cross-section;
right side – transverse diameter of the spinal cord cross-section

## 2.2. Discretization and mechanical properties

The white matter and grey matter were modelled as two different structures, using 8-node HEXA-HEDRA-type elements of three degrees of freedom in each node. Anatomical connection between them was made by common nodes at the contact areas. The strength parameters of the white and grey matter were set based on the experiments of ICHIHARA et al. [14]. The pia mater and dura mater were modelled using shell elements, with the parameters described by ŚCIGAŁA et al. [15]. The thickness of both the pia and dura mater were set based on the results of experiments of NICHOLAS and WELLER – 0.1 mm for the pia mater and 0.4 mm for the dura mater [16]. The denticulate ligament anatomically built as a composite structure was modelled using shell elements of the pia mater resistance parameters, covering the 8-node elements of HEXA-HEDRA core of the strength parameters of collagen [17]–[20]. A detailed description of the mechanical properties and finite elements used were presented in table 1 and figure 3. The example model of the cervical spinal cord is depicted in figure 4.

## 2.3. Model validation

Experimental validation was carried out based on a porcine spinal cord model. The main aim of the validation process was to verify the correctness of the model geometry plotting and discretization algorithms.

The spinal cord of a 6 month old pig was harvested from the spinal canal 3 hours after slaughter and placed in a 0.9% NaCl solution at 10 °C. Animals were sacrificed in the experiment conducted by another research team with the consent from the local ethics committee. The cervical region of the spinal cord was carefully dissected from the material obtained. Later we stripped the spinal cord of the dura mater and divided it into

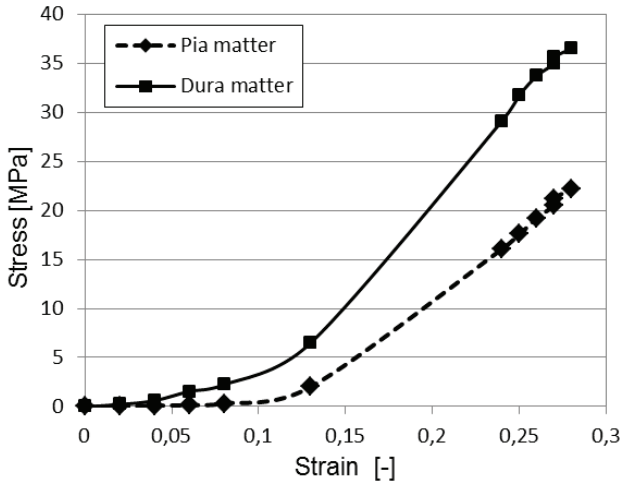


Fig. 3. Mechanical properties of pia and dura matter used in numerical model

Table 1. Mechanical properties of modelled anatomical structures with references and finite elements used. $E$ – Young modulus, $\nu$ – Poisson's ratio

| Material | Element | Mechanical properties | References |
|---|---|---|---|
| Grey matter | 8 nodes, brick | $E$ = 0.656 MPa, $\nu$ = 0.499 | [6] |
| White matter | 8 nodes, brick | $E$ = 0.277 MPa, $\nu$ = 0.499 | [6] |
| Pia mater | 4 nodes, shell | figure 3a | [15] |
| Dura mater | 4 nodes, shell | figure 3b | [15] |
| Denticulate ligament<br>a) Membranous component<br>b) Solid component | <br>a) 4 nodes, shell<br>b) 8 nodes, brick | <br>a) $E$ = 2.3 MPa, $\nu$ = 0.3<br>b) $E$ = 100 MPa, $\nu$ = 0.3 | <br>a) [17]<br>b) [25] |

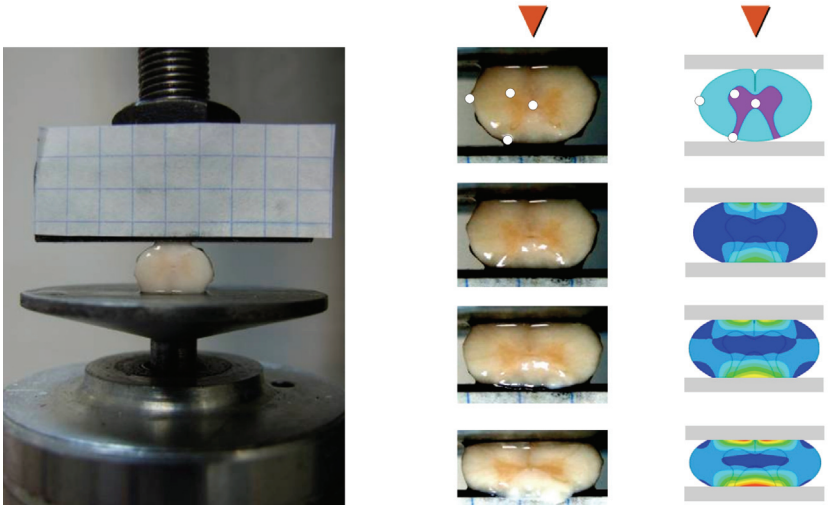

Fig. 4. The scheme of the experimental validation. Left side: experimental part of the FEM model validation. Progressive compression and deformation of the porcine spinal cord specimen are visible. Right side: progressive deformation of the FEM model under gradual compression. The white dots are indicating control points described in the text

seven anatomical segments. Each of the seven segments underwent dorso-ventral compression with the use of Material Testing System Synergie 100 (MTS Systems, Inc., USA). The compression process was documented photographically, and was carried out in a non-continuous manner, at 0.5-mm intervals until complete compression of the sample analyzed was reached. The axial section of the spinal cord segment was photographed before the commencement of compression (reference photograph) and after each subsequent compression by 1 mm. A millimetre scale was placed in the camera view. The following reference points were marked via computer assistance on the reference photographs of each spinal cord segment: 1) the place where the denticulate ligament is attached to the lateral surface of the spinal cord; 2) the location of the central canal; 3) the DREZ (dorsal root entry zone) location; 4) the centre of the anterior horn of the grey matter. The coordinates were also marked in a two-dimensional $X$, $Y$ system with the 0,0 point in the upper left corner of the standardized photograph. After localizing the reference points on the photographs of each compression, the coordinates were determined. This allowed the reconstruction of the movement of certain points during compression. Because the coordinates of the reference points were known, it was possible to determine precisely their distance from reference points at each compression stage.

A three-dimensional FEM model, comprising grey matter, white matter and the pia mater was consequently created for each sample based on the algorithm presented above. The geometry of each model accurately reflected the geometry of each segment of the spinal cord. A standard procedure described above was used to create each model. Strength parameters were also used according to the above mentioned scheme.

For every model each stage of compression was simulated using the FEM. The contact and sliding between the surfaces – virtual solid beams ($E$ = 200 GPa, $\nu$ = 0.3) and ventral and dorsal surfaces of the spinal cord, respectively, were modelled. Measurements of the location of reference points before and during compression were carried out. It was determined how the reference points moved at each stage of compression in relation to the points on the reference photograph. These

results were compared with the results obtained during the experimental phase of the study on the porcine spines. The results were analyzed statistically.

The scheme of the validation was presented in figure 4.

## 2.4. Calculation procedure

Calculations were carried out on XW 8600 workstations (Hewlett-Packard, USA) using the 11.0 and 12.1 versions of ANSYS Multiphysics software (ANSYS, Inc. USA). Each station was equipped with two four-core Intel Xeon X5470 3.33 64 bit processors, 16 GB RAM, and 300 GB RAID 0 HDD matrix.

## 2.5. Statistical analysis

The STATISTICA ver. 9.0 (StatSoft, Poland) was used to analyse the results of the study. Statistical significance was set at $p < 0.05$.

To analyse the normal distribution of certain features the Shapiro–Wilk test was used. To compare statistical significance of the validation of the mathematical model, the ANOVA Kruskal–Wallis test was used.

# 3. Results

A 3D FEM model of the human cervical spinal cord was developed. We found the procedure of the model generation to be simple and effective. Morphological properties of the FEM models corresponded well to the VHP images.

Sample images showing the FEM model are presented in figure 5.

Experimental model validation was performed according to the previously mentioned scheme. The experiment was repeated for four cervical spinal cord specimens. Therefore, the compression of 28 segments of the spinal cord made up the results. There were no statistically significant differences in the trajectories of the control points between the experimental and numerical models. The results of the Kruskal–Wallis ANOVA test, which evaluated the differences between the positions of the control points in numerical model and those from animal preparations at different stages of compression, are summarized in table 2.

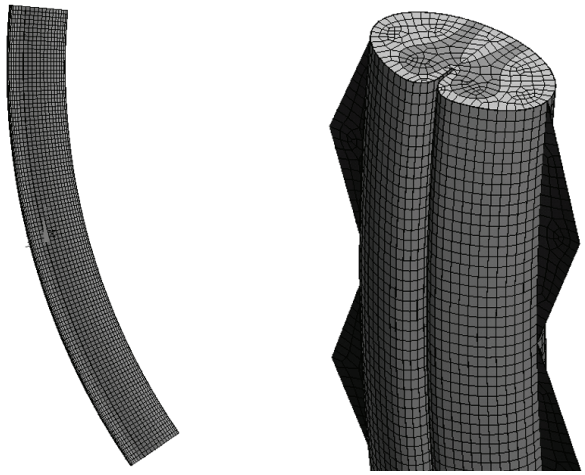

Fig. 5. Finite Element Model (FEM) of the human cervical spinal cord. Left side: lateral view of the dural sac. Cervical lordosis is noticeable. Right side: view of FEM model cross-section. Grey and white matter, dural sac and denticulate ligament are visible

# 4. Discussion

The traumatic spinal cord injury has been known from time immemorial. The first record from Edwin Smith's Papyrus from the 17$^{th}$ century BC says that the illnesses of this type cannot be treated. This opinion was strengthened throughout the next centuries and has lasted until the present times. The constantly developing knowledge of the pathophysiological mechanisms accompanying SCI helps to

Table 2. The results of the Kruskal–Wallis ANOVA test, which evaluated the differences between the position of the control points in numerical model and those from animal preparations at different stages of the compression. Stage I – from 0 to 1 mm, Stage II – from 1 to 3 mm, Stage III – from 3 to 5 mm compression. Values are presented as *X*, *Y* coordinates [*px*]

| | Experiment | | | | | | | | FEM | | | | | | | | *p* |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Control point | 1 | | 2 | | 3 | | 4 | | 1 | | 2 | | 3 | | 4 | | |
| Axis | *X* | *Y* | *X* | *Y* | *X* | *Y* | *X* | *Y* | *X* | *Y* | *X* | *Y* | *X* | *Y* | *X* | *Y* | |
| Stage I | 164 | 328 | 289 | 163 | 308 | 405 | 497 | 331 | 166 | 332 | 289 | 165 | 309 | 400 | 496 | 326 | 0.63 |
| Stage II | 144 | 344 | 258 | 203 | 304 | 410 | 498 | 361 | 146 | 348 | 258 | 205 | 305 | 405 | 497 | 356 | 0.98 |
| Stage III | 114 | 364 | 255 | 240 | 264 | 430 | 497 | 391 | 116 | 368 | 255 | 235 | 265 | 425 | 496 | 386 | 0.37 |

handle and implement a variety of experimental therapies [21].

In 2005, SHARMA published the results of a study conducted over many years on the function of the blood–spinal cord barrier in the pathophysiology of the primary and secondary spinal cord injury [2]. He showed that the damage to the blood–spinal cord barrier, which takes place in the first few seconds after the injury, is the starting point for the secondary spinal cord damage. It provokes the inflammatory response, induces oedema, leads to the formation of haemorrhagic and necrotic sites, and also leads to the activation of the apoptotic process.

Just two years later MAIKOS and SHREIBER showed that the extent of the damage to the blood–spinal cord barrier depends on the parameters of the mechanical stimulus acting on the spine at the time of injury [3]. In the FEM experiment, they proved that the induced mechanical strain correlated with the extent of the damage to the blood–spinal cord barrier [22], [23].

However, the idea of using advanced computer techniques in analyzing SCI first appeared in the nineties of the twentieth century [24]. Due to technical limits, the first professional attempts at using FEM in such analyses were undertaken in the twenty-first century.

In 2001 and 2003, ICHIHARA et al. [5], [14] reported the results of their valuable experiments in two-dimensional numerical models of the initiation of myelopathy during spondylosis. Despite the momentous importance of their research the clinical interpretation of the results obtained based on the analysis of two-dimensional biological structure models seems to be questionable.

A three-dimensional FEM model of the spinal cord was primarily used by WILCOX et al. in 2004 [7]. They attempted to simulate burst fracture of the thoracic spine, showing that the posterior longitudinal ligament plays an important role in SCI biomechanics and that the cerebrospinal fluid does not provide significant protection from sudden compression during SCI. However, this theory was verified in the paper of JONES et. al. four years later [25].

The first published attempt to use a three-dimensional model for clinical analysis of SCI was made in 2008 [26]. A successful simulation of SCI was run in a patient, whose prognosis on the extent of improvement of the neurological state based on the FEM model was correlated with the clinical examination and imaging conducted six months after the injury. Due to the fact that the model used has not undergone the process of experimental validation it was then suggested that such studies need to be continued.

Since then, three major studies have been published concerning the related topic [1], [9], [27]. Although these studies were not based on a clinical group, they dealt with the theoretical aspect of the available knowledge of this matter. All three models presented were designed as three-dimensional solids, whose geometry was based on published anatomical data. GREAVES et al. [1] and LI et al. [27] decided to create the model of the spinal cord with white and grey matter using tetrahedral elements. This seemed justifiable, but may have negatively influenced the accuracy of the results achieved. In both above mentioned articles, the validation process described was based solely on data available from literature – the authors of these articles did not calibrate or validate their models experimentally.

The previously published values of the strength characteristics of the dura and pia mater of the spinal cord allowed the modelling of these structures based on a non-linear stress–strain characteristics [15]. They are typical of biological objects and have been previously described, but not accurately enough that could allow them to be used for numerical modelling [28]. We decided to use the author's data obtained experimentally [15].

The unique feature of the model is its possibility of interaction with the operator at an early stage of preparing its components. The possibility of creating topographical correlations between the white and grey matter based on the sagittal and transverse measurements of the patient's spinal cord seems to be particularly useful and valuable.

The use of virtual denticulate ligaments to secure the spinal cord in the three-dimensional spinal cord space is also innovative. Both its ends are fixed, as is in real spinal cord, but it also ensures a quasi-physiological stabilization in the antero-posterior axis [4]. Although the model lacks cervical vertebrae, this cannot be considered as a clear drawback. At the present stage of studies, the lack of vertebrae is compensated by the fixation and deformation of the dura mater and the modelling of the contact between the dura and pia mater covering the spinal cord. The mechanics of the osseous and ligamentous structures of the spine can play a role in dynamic analyses. Therefore future studies of this topic should be carried out.

Our validation process consisted in observing the controlled compression of the porcine cervical spinal cord specimens and in carrying out the experiment in the FEM environment. It is a widely known fact that post-mortem autolysis leads to rapid changes of me-

chanical properties of the soft tissue [14]. For this reason we decided to use porcine spinal cords specimens instead of human ones. Animal tissues could be obtained and preserved under optimal conditions, which, for obvious reasons, was not possible for human specimens. Additionally, as shown in the literature, the porcine spinal cord should be just as suitable as human specimens in biomechanical testing [23], [25].

The development of an experiment that could be reconstructed using FEM allowed for an observation of the behaviour of the model under controlled conditions. The validation was successfully completed. It is, however, worth noting that the loads applied to the spinal cord were static loads. This could determine the implementation of the quasi-dynamic method of the FEM analysis, which is based on pushing virtual bony fragment into the spinal cord with an infinitely high speed. This method is considered effective and is used in similar simulations [1], [27]. However, it prevents the evaluation of speed as a factor influencing post-injury changes. Taking into consideration the intended broadening of the experiment, which should include carrying out dynamic analyses, an experimental validation of the FEM model including velocity seems to be crucial.

The authors of the study showed that numerical modelling with the use of FEM might be applicable in the analysis of SCI in humans. The results of such an evaluation may be useful not only immediately after the injury – in the phase of deciding on the therapeutics used to treat the patient, but may also be used to create a prognosis of the neurologic state of the patient after a specific period of time.

# 5. Conclusions

It is possible to simulate a traumatic injury to the cervical spinal cord with the use of our validated numerical FEM model. Clinical validation is crucial for the further studies and should be carried out on a large group of patients.

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