A finite element study of traditional Chinese cervical manipulation

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Received: 29 December 2016 / Revised: 7 June 2017 / Accepted: 11 June 2017 / Published online: 28 June 2017
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Abstract

Purpose Traditional Chinese cervical manipulation (TCCM) has been claimed as an effective treatment for diseases of the cervical spine, but its biomechanical effects on the vertebral body and intervertebral discs remain unclear. The aim of this study was to develop and validate a detailed finite element model of cervical spine, which was then used to investigate the biomechanical response of the cervical spine to TCCM.

Methods The model of a C2–T1 cervical spine was constructed based on CT images of a healthy male volunteer and validated against published in vitro studies under different loading conditions. The detailed force–time data of TCCM were measured on the same volunteer through dynamometric diaphragms. The data were applied on the validated finite element model to simulate TCCM.

Results The current model could offer potentials to effectively reflect the behavior of human cervical spine suitable for biomechanics studies of TCCM. Under simulated TCCM condition, the stress distributions in cervical spine and intervertebral discs could not be completely explained through the traditional theory.

Conclusion Spinal manipulation, or TCCM, might play no role in reducing intradiscal pressure for treating cervical spondylosis. It could cause less stress concentration in intervertebral discs while operating spinal manipulation or TCCM when the adjustment points was chosen near the root of spinous process than the top of spinous process.

Keywords Finite element analysis · Cervical spine · Biomechanics · Manual therapy · In vivo measurement

Introduction

Manual therapy is one of the recommended treatments in managing neck pain without presence of neurologic signs (Neck Pain and Associated Disorders grade I or II) [1]. Besides its effectiveness, some adverse events (e.g., cervical arterial dissection and disc rupture) and erroneous beliefs about the cracking sound during such treatment have been reported recently, which raised concerns of safety about manual therapy [2, 3]. Traditional Chinese cervical manipulation (TCCM) is a long-established characteristic of traditional Chinese massage (referred to as tuina). It is believed that neck symptoms result from channel blockage and joint displacement, and TCCM may clear the channels and utilize joint manipulation to restore joint alignment [4]. At present, this technique is widely accepted as a complementary and alternative medicine modality [5]. Although this manipulation is routinely used...
to treat cervical spine diseases like other manual therapy, the effectiveness of this therapy remains controversial. Lin et al. [6] reported that TCCM was effective in treating neck pain patients, but a systematic review reported that there was limited evidence indicating that TCCM could produce short-term improvement for neck pain [4], and some authors considered it to be a risky practice [7]. In addition, the biomechanical effects on the intervertebral disc during TCCM remain inconclusive [8]. One study investigated the stresses and displacements in cervical nuclei pulposi during TCCM [9], but its methodology was not well introduced. It is hard to understand TCCM without quantitative biomechanical information.

Due to ethical considerations and the limitations in experimental equipment and techniques, very little biomechanical information can be obtained through in vivo trials. It is also difficult to simulate physiological and pathological behaviors with in vitro experiments. In this instance, computational simulation can provide a satisfactory option for gathering such biomechanical data [10, 11]. As a representative computational method, finite element (FE) modeling can also provide insight into the inner workings of the cervical spine, and can provide detailed biomechanical information of the human neck in intact, injured, or stabilized states [11–13].

This study aimed to develop an FE model of cervical spine to intimately examine the cervical spine during the course of TCCM. To validate the model against those given in literature, different types of loadings were applied to the model under conditions of flexion, extension, right and left bending, and right and left axial rotation. The movements and loadings simulated here followed recognized TCCM cervical manipulation techniques.

Materials and methods

FE modeling

A healthy Chinese male volunteer (age 30 years, height 170 cm, body weight 68 kg) was recruited for this study. The subject’s skull and vertical spine were scanned using a CT scanner (Brilliance 64, Philips Electronics, Netherlands). The final CT images had a resolution of 0.54 mm × 0.54 mm and the slice interval of 0.625 mm. This study protocol was approved in advance by the Medical Ethics Committee of University.

A detailed three-dimensional (3D) non-linear FE model of an intact C2–T1 spine was created based on the transverse CT images. Within the software Mimics 17.0 (Materialise Inc., Leuven, Belgium), these images were segmented and translated to various 3D solid volumes of all vertebrae. In this software, the height and width of each vertebral body, as well as the anterior and posterior thicknesses of the discs were measured. These parameters then were compared with data from the literature [14]. Finally, the solid volumes were created to fill the spaces between the vertebrae to create intervertebral discs. The final constructs were exported as STL format files (Fig. 1a). The solid volume was then, respectively, imported into the software Geomagic Studio 12.0 (Geomagic Inc., USA), in which it was converted into a non-uniform rational B-spline surface geometry structure.

Each intervertebral disc was modeled as a central nucleus surrounded by an annular ground substance reinforced by fibers acting at approximately ±25° from the transverse plane [15, 16]. The annulus also accommodated the fiber definitions in each element. The nucleus was modeled as an incompressible fluid, whose volume was approximately 33% of the entire disc volume [17]. Five major cervical spine ligaments were incorporated into the model: anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), ligamentum flavum (LF), interspinous ligament (ISL), and capsular ligaments (CL). The origin, insertions and numbers of the ligaments connected to vertebrae were determined from anatomical texts and related literature [18–21]. The ligaments were modeled as 3D truss elements acting nonlinearly in tension only via a hypoelastic material designation in ABAQUS 6.13 (Abaqus Inc., USA).

To reduce the resources required for creating a mesh of the complex spinal geometry, ABAQUS was used to generate a tetrahedral mesh on the vertebrae and a hexahedral mesh on the discs (Fig. 1b). The material properties of the various tissues used in the model were derived from literature [12, 22] and are listed as Table 1.

Model validation

To validate the FE model, pure moment loads were applied to the model in order to simulate flexion, extension, lateral bending and axial rotation through three anatomical planes. For flexion and extension, the loads of 0.5, 1, 1.5 and 2 N m were, respectively, applied. For both lateral bending and axial rotation, the load was 1 N m. For each simulation, the load was applied to a flying-node that was created 1 mm above the odontoid process of C2. The nodes lying on the odontoid process of C2 were coupled with the flying-node to create a pure moment. All coupled nodes were constrained in all directions. The range of motion (ROM) of each adjacent vertebra under different loading conditions was then calculated [23, 24].

The results of the FE model under pure moment loads were compared against a number of in vitro studies [24–29] conducted under different loading planes in flexion/extension, right/left lateral bending, and right/left axial rotation.
Fig. 1 Three-dimensional model of intact cervical vertebra (C2–T1): a solid model; b finite element model

Table 1 Material properties and element types used for various components in the current model

<table>
<thead>
<tr>
<th>Component</th>
<th>Element type</th>
<th>Young’s modulus (MPa)</th>
<th>Poisson’s ratio</th>
<th>Cross-sectional area (mm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bony structures</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Vertebral cortical bone</td>
<td>Isotropic, elastic tetra element</td>
<td>10,000</td>
<td>0.3</td>
<td>–</td>
</tr>
<tr>
<td>Vertebral cancellous bone</td>
<td>Isotropic, elastic tetra element</td>
<td>450</td>
<td>0.23</td>
<td>–</td>
</tr>
<tr>
<td>Posterior bone</td>
<td>Isotropic, elastic tetra element</td>
<td>3500</td>
<td>0.25</td>
<td>–</td>
</tr>
<tr>
<td>Intervertebral disc</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Annulus (ground)</td>
<td>Neo-Hookean, hex element</td>
<td>4.2</td>
<td>0.45</td>
<td>–</td>
</tr>
<tr>
<td>Annulus (fiber)</td>
<td>Rebar</td>
<td>450</td>
<td>0.3</td>
<td>–</td>
</tr>
<tr>
<td>Nucleus</td>
<td>Incompressible fluid element</td>
<td>1</td>
<td>0.49</td>
<td>–</td>
</tr>
<tr>
<td>Ligament</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>ALL</td>
<td>Tension only, truss elements</td>
<td>15(&lt;12%) 30(&gt;12%)</td>
<td>0.3</td>
<td>11.1</td>
</tr>
<tr>
<td>PLL</td>
<td>Tension only, truss elements</td>
<td>10(&lt;12%) 20(&gt;12%)</td>
<td>0.3</td>
<td>11.3</td>
</tr>
<tr>
<td>LF</td>
<td>Tension only, truss elements</td>
<td>5(&lt;25%) 10(&gt;25%)</td>
<td>0.3</td>
<td>46</td>
</tr>
<tr>
<td>ISL</td>
<td>Tension only, truss elements</td>
<td>4(20–40%) 8(&gt;40%)</td>
<td>0.3</td>
<td>12</td>
</tr>
<tr>
<td>CL</td>
<td>Tension only, truss elements</td>
<td>7(&lt;12%) 30(&gt;12%)</td>
<td>0.3</td>
<td>42.2</td>
</tr>
<tr>
<td>Joint</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Facet (apophyseal joint)</td>
<td>Sliding surface to surface contact</td>
<td>–</td>
<td>–</td>
<td>–</td>
</tr>
</tbody>
</table>

ALL anterior longitudinal ligament, PLL posterior longitudinal ligament, LF ligamentum flavum, ISL interspinous ligament, CL capsular ligament
The predicted ROM for each motion segment was computed and compared with the experimental data. Furthermore, the results were compared with a recent FE model of the full cervical spine reported by Panzer et al. [30]. For flexion and extension, the loads applied to the model were 0.5, 1, 1.5, and 2.0 N m. These parameters were in line with studies by Nightingale et al. [26, 27], while Wheeldon et al. [25] and Panzer et al. [30] only used a load of 0.3 N m. For lateral bending and axial rotation, the load was 1.0 N m in all experiments [24, 28, 29]. Generally, the results of validation would be accepted as good agreement when the calculated ROM was within the range of mean value ± one standard deviation of in vitro measurement.

**TCCM simulation**

The TCCM simulation consisted of two steps. Firstly, we measured in vivo loading parameters from a volunteer during manipulation process. Then, the recorded force was applied to the model.

For in vivo measurement of TCCM, the volunteer mentioned above also accepted a TCCM cervical manipulation of the low cervical spine from C4 to T1. The operator is a clinical specialist (Doctor of traditional Chinese medicine) with 20 years experience dealing with cervical spine diseases using manipulative therapy and other non-operative treatments. The force–time diagram recorded during the operation processing and the definitions for the preload force, peak force and the three phases are shown as Fig. 2. The most important step of TCCM is the adjustment step, which contains a preload phase, thrust phase and resolution phase [31]. In order to acquire the adjustment force applied to the neck of volunteer, dynamometric diaphragms and a Runinsense force measurement system (Walkinsense sports, Tomorrow options, UK) were used. To measure the pressure on the volunteer’s neck, one diaphragm was pasted on the spinous process of C5, and the other was pasted on the right tuberculum mentale. These two locations were the main loading points during treatment. The forces experienced during the adjustment step, particularly the peak force, were extracted to calculate the moment to be applied to the FE model.

The whole TCCM simulation was performed under three different loading conditions:

1. **Pure head-weight loading (PH)** The mass of the volunteer’s head was estimated from the CT image data. A 55-N preload was applied to the superior faces of C2 to simulate the weight of the head. This condition simulated the mechanical state of cervical spine in the pre-manipulation position.

2. **Adjustment force loading on the top of the C5 spinous process (AD1)** The peak force was applied perpendicular to the axis of spinous process of C5 along the horizontal plane of cervical spine.

3. **Adjustment force loading on the root of the C5 spinous process (AD2)** The direction and magnitude of force was the same as described in loading condition 2.

During FE analysis of TCCM, all the nodes on the inferior endplate of T1 and superior endplate of C2 were rigidly fixed to simulate the pro-orientation of the patient. The FE simulations were performed in the static mode using ABAQUS standard solver.

**Results**

**FE modeling and validation**

Figure 3 shows ROM comparisons under flexion and extension of this FE model against similar in vitro and computational studies. The flexion ROM of C2–C3, C5–C6, and C6–C7 was less in the current model than the mean values reported by Wheeldon et al. [25], while the C3–C4 and C4–C5 motion curves showed good correlation to the mean experimental values. This simulation also had similar flexion ROM outcomes to a study by Panzer et al. [30].

In extension, the model produced comparable results to those reported by Wheeldon et al. [25] in the C6–C7 segment. However, in other segments, the FE model showed up to 49% greater extension. Compared to the result by Panzer et al. [30], the FE model here showed a greater extension ROM. In C6–C7, both models predicted nearly the same rotation response to the study by Wheeldon et al. [25].
Plots of the full ROM in lateral bending and axial rotation of various segments are presented as Fig. 4. Under lateral bending, the FE model in this study produced lower values than the mean values of Panjabi et al. [24] and Moroney et al. [28]. However, the current model predicted that the maximum rotation would occur in C2–C3 (8.3°), while the lowest rotation would occur in C6–C7 (4.1°). These maximum and minimum values were comparable to those reported by Panjabi et al. [24] and Moroney et al. [28].

In axial rotation, all results were less than those of Lysell [29]. All segments, except C2–C3, showed a similar trend to Panjabi et al. [24] and Lysell [29]. The maximum ROM occurred in C4–C5, followed by C3–C4 and C5–C6. The minimum rotation occurred in C6–C7. In the C2–C3 segment, the current FE model was more flexible than that
of Panjabi et al. [24], and a little stiffer in the C4–C5 segment.

**TCCM simulation**

The von Mises stress distributions around the vertebrae and intervertebral discs are presented in Fig. 5. Under PH loading, the main stress concentrations were around the posterior border of the vertebrae, particularly at the lamina and pedicle of T1 and ISL. In the intervertebral discs, the posterior fibrous rings experienced the highest stress.

The internal von Mises stress of the whole adjustment simulation (AD1 and AD2) showed the main stress concentrations in the simulation of PH. The C5 upper surface bore the weight of the head, while the inferior surface of C5 shared the head weight and adjustment force. The stress concentrations showed its similarity and difference when simulating the adjustment of the C5 spinous process on different adjustment points.

In the simulation of AD1 and AD2, the high-stress region appeared in the lamina and spinous process of C5 as well as the ipsilateral ALL, and facet joints of C5–C7. The stress on the vertebrae gradually decreased from C5 to T1, and the posterior regions of the intervertebral discs shared the highest stress. In addition, the right discs experienced much greater stress than the left side. Some element deformations could even be observed on the C5–C6 intervertebral discs. The most difference between AD1 and AD2 came to the stress value. The stress value decreased a lot in the right ALL, left facet joints and the intervertebral discs of C5–C7.

**Discussion**

In the validation step, the model was compared against several in vitro experimental studies with a wide loading range. In general, there was good agreement with these referenced experimental studies, which indicated that the current FE model can effectively reflect the motions of a human cervical spine. However, mostly in extension, the model was found to be much more flexible than these in vitro studies, particularly between C2 and C5. In addition, the cervical spine was found to be stiffer in extension than in flexion. Similar findings have also been reported elsewhere [25–28]. It has been suggested that such lack of continuity between datasets could be attributed to differences in subject age, fixation method, and boundary conditions [26]. Anatomically, facet joints are the major components for bearing loads and play an important role in extension movements. Therefore, variations in the facet joint orientations and initial positioning may be a more significant factor in explaining the difference. On the other hand, the human neck is asymmetric in the sagittal plane, with resulting asymmetric activation of ligaments.

In axial rotation, each segment of this FE model was stiffer than the ROM values reported by Lysell [29]. However, the ROM of C2–C3 was within the limits of similar studies [24, 29], but it was also greater than that reported by Moroney et al. [28]. In the C6–C7 unit, the rotation ROM was comparable to referenced studies [24, 28]. However, Moroney et al. [28] reported that segments with the same ROM, but their method violated the continuity of longitudinal structures [24].

Simulations were conducted under three loading conditions. Under PH loading, higher stress was mainly distributed on the lamina and pedicle of T1, as well as ISL from C2 to T1. This is likely due to the fact that the endplate of T1 was rigidly secured to ensure a fixed base and due to the slight flexion of the cervical spine after loading to 55 N. Stresses in the discs were concentrated at the outer annulus fibrosus, while the nucleus pulposus region shared very low stress.

In the AD1 and AD2 simulation, when the adjustment force was loaded on the model, the stress in the superior
vertebrae, ISLs and discs of C5 reduced compared with the original model. The greatest stress appeared in the lamina and spinous process of C5 as well as ipsilateral ALL and facet joints of C5–C7. The left facet joints of C5–C6 also showed a high-stress region, which was mainly caused by the tension from CL to resist the adjustment force. Furthermore, the stress was significantly increased in C5–C6 and C6–C7 discs.

Fig. 5 Internal von Mises stress of the anterior and posterior view of spine, and internal von Mises stresses in the discs for different simulation conditions
In comparison to AD1 and AD2, the high-stress regions were similar to each other except the stress value in the same elements. The AD1 loading condition produced greater stress in the lamina and spinous process of C5, C6 and T1, as well as the upper ISLs of C5. The stresses in the right ALL and the facet joints of C5–C7 were distinctly greater in this model than under the PH and AD2 loading conditions. Under the effect of adjustment force, the low stress regions moved to left while the high-stress regions moved to right in the direction of the adjustment force. For the intervertebral discs, the stress patterns observed in the upper discs of the C5 segment were very similar to those seen in the PH simulation.

A traditional viewpoint holds that intradiscal pressure can be reduced through spinal manipulation [32], but this was not supported in this study. On the contrary, the high-stress regions increased in the disc while an adjustment force was applied. This may indicate that spinal manipulation, or TCCM, might play no role in reducing intradiscal pressure for treating cervical spondylosis. One study reported that most participants had erroneous beliefs about the cracking sound produced during spinal manipulation, e.g., discs return to their normal position [3]. In the current study, since we found that TCCM could not reduce intradiscal pressure in an FE model, it is not likely that a herniated disc would be pushed back into its original place.

AD1 and AD2 were the mainly two adjustment points when operating TCCM in China [6]. The maximum von Mises stress value decreased from 16.28 to 10.11 MPa in the right-back element of C5–C6 disc. The same phenomenon appeared in the right ALL and left facet joints of C5–C7. It might be caused by a shorter lever while operating AD2 than AD1. This phenomenon suggested that if TCCM is necessary while treating cervical diseases, the adjustment points near the root of spinous process would be a better choice than the top of spinous process.

One limitation of computational simulation is the model simplification, which often makes it difficult to directly compare against in vitro data. The material properties incorporated in various components of the spine was derived from literature, and then be further simplified. In the current study, the anatomical data were collected from a healthy 30-year-old male for constructing the FE model. The geometry was subject-specific, and any degeneration or abnormal morphology was not considered. Factually, the age and gender were reported to influence the lordosis value and intervertebral disc height of cervical spine [33]. These factors would potentially cause variation on the results in the validation process and TCCM simulation. Therefore, the results of the current study could only be applied in male subjects about similar age. Future studies are required to consider the aforementioned factors for more general knowledge.

Clinically, some case studies reported disc herniation after cervical manipulation [34, 35], which showed the same concern as the current study did, about the mechanism on the disc during manipulation. In the present study, the intervertebral discs were assigned with normal material property and geometry. However, the degeneration would occur with aging and may affect the intervertebral disc height, geometry and the material property [36]. Further studies concerning factors such as disc degeneration should be carried out to clarify which groups of subjects may be at higher risk of injury for manipulation therapy. Addressing these issues would enhance the safety of such treatment. On the other hand, although the current study suggested that TCCM might involve no contribution to reduce intradiscal pressure, the clinical effectiveness of cervical manipulation may be achieved by stimulating receptors within deep intervertebral muscles [37]. Thus the tension in the deep segmental muscles should also be further studied to explore the potential mechanism of TCCM.

Additionally, the simulations were performed under idealized conditions that did not consider muscle forces and failed to include a number of ligaments of the neck. The adjustment carried out during TCCM was also decomposed and simplified to effectively incorporate it into the FE model. A more detailed and accurate FE model will be constructed in future studies.

Conclusion

A detailed FE model of the C2–T1 cervical spine was developed and validated to simulate TCCM or spinal manipulation. It suggested that TCCM or spinal manipulation could significantly alter the stress regions around cervical vertebrae, both in location and value, but may make no direct contribution to reducing intradiscal pressure. The adjustment points near the root of spinous process would be a better choice when the treatment is really needed.

Compliance with ethical standards

Conflict of interest The authors declare no competing interest.

Ethical approval All procedures performed in studies involving human participants were in accordance with the ethical standards of the institutional and/or national research committee and with the 1964 Helsinki declaration and its later amendments or comparable ethical standards.

Informed consent Informed consent was obtained from all individual participants included in the study.
References


