Clinical Biomechanics 28 (2013) 846–852

Contents lists available at ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

Effect of the transverse ligament rupture on the biomechanics of the cervical spine under a compressive loading

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A R T I C L E   I N F O

Article history:
Received 8 October 2012
Accepted 24 July 2013

Keywords:
Cervical spine
Transverse ligament rupture
Compressive load
Extension moment
Flexion moment
Finite element
Atlas–dens interval
ADI
Space available of the cord
SAC
Kinematics
Contact force

A B S T R A C T

Background: In order to diagnosis a transverse ligament rupture in the cervical spine, clinicians normally measure the atlas–dens interval by using CT scan images. However, the impact of this tear on the head and neck complex biomechanics is not widely studied. The transverse ligament plays a very important role in stabilizing the joint and its alteration may have a substantial effect on the whole head and neck complex.
Methods: A finite element model consisting of bony structures along with cartilage, intervertebral discs and all ligaments was developed based on CT and MRI images. The effect of head weights (compressive load) of 30 N to 57 N was investigated in the cases of intact and ruptured transverse ligament joints. The model was validated based on experimental studies investigating the response of the cervical spine under the extension–flexion moment.
Findings: The predictions indicate a significant alteration of the kinematics and load distribution at the facet joints of the cervical spine with a transverse ligament tear. The vertebrae flexion, the contact force at the facets joints and the atlas–dens interval increase with the rupture of the transverse ligament and are dependent to the head weight.
Interpretation: A transverse ligament tear increases the flexion angle of the head and the vertebrae as well as the atlas–dens interval. The atlas–dens interval reaches a critical value when the compressive loading exceeds 40 N. Supporting the head after an injury should be considered to avoid compression of the spinal cord and permanent neurologic damage.

1. Introduction

A cervical spine can be subjected to severe injury through various types of accidents and/or extreme movements. Motor vehicle accidents and falls are common causes. However, many other causes exist that could result in a fracture and/or injuries to the cervical spine structures, bone and soft tissues, such as athletics and activities with violent physical contact or without contact. The rupture of the transverse ligament (TL) is one of these injuries caused by a fall with a blow of occiput involving an anteriorly directed force to the back of the head with forced flexion (Hasharoni and Errico, 2005) and resulting in an excessive relative anterior displacement of C1 against C2 (Fielding et al., 1976; Hasharoni and Errico, 2005). The rupture of the TL occurs typically in the body of the ligament; however it may occur also at the bone tubercle (Fielding et al., 1974). The TL in the cervical spine is recognized as the primary structure stabilizing the cranio-vertebrae joint (Fielding et al., 1974; Heller et al., 1993) restricting both the flexion and the anterior displacement of the atlas (Dvorak et al., 1988). In their mechanical and histological study, Dvorak et al. (1988) measured the breaking strength of the alar ligament and the transverse ligament to be 200 N and 350 N, respectively. The atlas-dens interval (ADI), defined as the distance between the anterior portion of the atlas and the dens of the axis, exceeding 3 mm for an adult was found to be an implication of either or both the transverse ligament (Dreyer and Boden, 1999; Fielding et al., 1974; Hein et al., 2002; Kontautas et al., 2005; Lee et al., 1997; Schären and Jeanneret, 1999) and the alar ligament (Fielding et al., 1974) failure. For others, the ADI should not exceed 3 to 3.5 mm (Hasharoni and Errico, 2005). In fact, the ADI is often used to assess trauma to the cervical spine based on its value increase that can be an indication of the TL rupture. The ADI increase results in posterior displacement of the dens relative to the atlas and consequently reduces the space available for the cord (SAC), recognized also by the Posterior atlanto-dental interval (PADI), which in turn consequently compresses the spinal cord (Oda et al., 2009). Race et al. (2008) reports also that a quick and reliable identification of unstable fractures of this region is a must to avoid death and disability after trauma. Surgical treatment with posterolateral decompression and posterior C1–C2 fixation was proposed to avoid compression of the spinal cord (Takeuchi et al., 2011). Most of the investigation of TL failure deals with the determination of the ADI by using CT scan and MRI images. However, biomechanics of the head and neck complex as well as the determination of the effect of the rupture of the TL on the joint is not widely studied. For this reason, the aim of this present study is to evaluate the effect of the TL rupture on the biomechanical response of the head and neck complex under a static compressive loading varying from 30 N to 57 N.

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0268-0033/$ – see front matter © 2013 Elsevier Ltd. All rights reserved.
http://dx.doi.org/10.1016/j.clinbiomech.2013.07.016

To achieve this, a finite element (FE) head and neck complex model was developed based on CT scans and MRI images of a 39 year old male by including the bone structures, intervertebral discs, and all ligaments and cartilages. Then, the response of the developed model was validated and compared with experimental studies existing in the literature under the flexion and extension moments loading applied to the center of mass of the head (Panjabi et al., 2001 and Wheeldon et al., 2006). Thereafter, the effect of the TL rupture on the biomechanics of the head and neck (HN) complex was investigated under a static compressive load. The compressive load can be seen as the variation of the head weight of the population. In this study, we assume that the muscles are not activated which leads to their absence in this model. We hypothesized that the TL rupture influences the biomechanical response of the cervical spine and the amount of this influence depends on the amount of compressive loading.

2. Methods

A 3D nonlinear HN complex FE model was developed and constructed based on CT scans and MRI images of a 39 year old male. The images were provided by the National Library of Medicine's Visible Human Project; the axial CT scan images are spaced 1.0 mm apart. Reconstruction of the head, cervical vertebrae (C1–C7) and the first thoracic vertebra (T1) was based on CT scan images and reconstruction of the intervertebral discs was based on MRI images. Reconstruction was done using Mimics software (Materialise N.V., Leuven, Belgium, Version 12.01).

The cartilages of the Atlas (C1), the Dens (C2) and the facet joints were reconstructed with Abaqus FE software (Dassault Systèmes Simulia Corp., version 6.7), and modeled with shell elements due to their thinness compared to the global size of the cartilaginous structure. As well, ligaments were modeled with Abaqus FE software and their insertions were obtained from anatomic and histological studies (Panjabi et al. (1991a,b); Primal Pictures 3D, 2007; Yoganandan et al., 2001).

The FE model consists of bony structures (head, cervical vertebrae (C1–C7) and the first thoracic vertebra (T1)) and their cartilage facet joints, intervertebral discs and all ligaments. The considered ligaments are: alar ligaments, apical ligament, anterior longitudinal ligaments (ALL), capsular ligaments, flavum ligaments (FL), posterior longitudinal ligaments (PLL), transverse ligament (TL), interspinous ligaments, intertransverse ligaments, supraspinous ligaments, anterior atlanto-occipital membrane (AAOM) and the posterior atlanto-occipital membrane (PAOM). All the ligaments were individually modeled through a number of nonlinear uniaxial spring elements except for the TL which was modeled as a bundle (Shell elements) with nonlinear material properties (Fig. 1b). The stress–strain curves of the Alar ligament, ALL, PLL, TL, Interspinous Ligament, FL and capsular ligament were obtained from the slow rate measurements of Shim et al. (2006). All the stress–strain curves are presented in Fig. 2. The considered cross-section areas of the ligaments are summarized in Table 1 (Dvorak et al., 1988; Shim et al., 2006; Yoganandan et al., 2001). In this study, the bony structures were considered as rigid surfaces, meshed by shell elements, due to their greater stiffness than the adjacent soft tissues and the considered loading. However, the intervertebral discs, each consisting of an annulus and nucleus, were modeled with hexagonal solid elements with homogenous elastic material property (Dauvilliers et al., 1994; Kleinberger, 1993; Nitsche et al., 1996; Yang et al., 1998; Yoganandan et al., 1995; Yoganandan et al., 1996; Zhang et al., 2006). The annulus and nucleus each have a Young's modulus of 7 MPa and a Poisson's ratio of 0.4 and 0.49, respectively (Yoganandan et al., 2001). Iatridis et al. (1996) and Pooni et al. (1986) found that the inner annulus fibrosis cross-section area measures between 30 and 60% of the total cross-sectional area of the non-degenerate intervertebral disc. In the developed model, each intervertebral disc computer-aided design (CAD) model was partitioned to create the annulus and nucleus regions assuring an interaction by shared nodes at the common surface between the two regions (Fig. 1b). The partition was done so that the cross section area of the inner annulus fibrosis is at approximately 50% of the total cross-section area of the intervertebral disc.

The cartilage layers in the whole joint are considered as shell with homogenous and isotropic material. The considered thickness of each cartilage is 0.5 mm based on the measured mean value in a study by Womack et al. (2008). A Young's modulus of the cartilage of 10 MPa and Poisson's ratio of 0.45 were considered in this study. To define the location of the center of mass of the head (CMH), we used the averaged position found by Walker et al. (1973) using X-rays.

In the first section of this study, we intend to validate the developed FE model, pure moments of extension (−1.0 N.mm) and flexion (+1.0 N.mm) were applied to the CMH. For stable unconstrained boundary conditions, T1 was fixed while the cervical vertebrae and head were left free. The results will be compared to the experimental studies found in the literature for the same nature of movement.

In the second section of this paper, the effect of the rupture of the TL under a compressive load (CL) was investigated. To achieve this objective, two cases were considered; the first one considers an intact TL; however the second one simulates a case with a ruptured TL. A compressive load varying from 30 N to 57 N was applied to the CMH and was considered for the analyses. Mean head weight was measured to be 40 N (Hodgson et al., 1970) and the considered head weight can be seen as the upper and lower limit of the head weight of a sample of the population. In addition, the considered head weight greater than 40 N can be considered as a normal head that supports a supplementary weight like a helmet. The joint reference configuration for each studied case was initially established by the stabilized position before applying any loads. The Abaqus FE software convergence tolerance default criteria were used for FE analyses in this study.

3. Results

3.1. Model validation

Under a flexion moment applied to the CMH, the cervical spine vertebrae and the head move anteriorly, distally and flex (Fig. 3a–d). The absolute angle of flexion calculated according to T1 increases from C7 to the head. At 1.0 N.m flexion moment, the angle of flexion of the head and C1 is 14.3° and 12.0°, respectively (Fig. 3a). The relative angle of flexion calculated according to the distal adjacent vertebra is greater in the cephalic cervical levels (C0–C1 and C1–C2) than in the caudal vertebrae level (Fig. 3e and f) and the maximum computed angle of flexion was at the C1–C2 level (Fig. 3e). The relative angle of flexion or extension is greater when the moment increases. However, this relative angle, which is an index of flexibility, is not necessarily dependent on the proximal–distal location of the vertebrae. For example, the most flexible vertebral level is C1–C2 then C0–C1 then C3–C4 whether under the flexion or extension moments and then for C6–C7 under the flexion moment and C2–C3 level for the extension moment loadings (Fig. 3e and f).

The proximal–distal displacement of the head and cephalic vertebrae is sensitive to the magnitude of the applied moment. A 1.0 N.m flexion moment induces a 3.6 mm distally for the head, 0.1 mm for C1 and 0.5 mm proximally for C2. On the other hand, an extension moment of 1.0 N.m moment induces only a distal displacement with a maximum of 2.1 mm computed at C1 (Fig. 3b).

Under flexion and extension moment loading cases, the head and vertebrae move anteriorly and posteriorly, respectively (Fig. 3c and d). The magnitude of these translations depends on the proximal–distal location of the vertebrae; it is greater as the vertebra is proximal. Under the 1.0 N.m flexion moment, the anterior displacement of the head and C1 reaches 20.4 mm and 11.2 mm, respectively (Fig. 3c). Note that the lateral displacement coupled with the considered loadings is very small and its maximum does not reach 0.8 mm at the head.

The contact force (CF) at the facet joints increases with an increase in the applied moment for C0–C1 and C1–C2 joints. The CF is zero at C2–C3