Subject-Specific Inverse Dynamics of the Head and Cervical Spine During *in Vivo* Dynamic Flexion-Extension

William J. Anderst¹
e-mail: anderst@pitt.edu

William F. Donaldson, III

Joon Y. Lee

James D. Kang

Department of Orthopaedic Surgery,
University of Pittsburgh,
Pittsburgh, PA 15203

The effects of degeneration and surgery on cervical spine mechanics are commonly evaluated through in vitro testing and finite element models derived from these tests. The objectives of the current study were to estimate the load applied to the C2 vertebra during in vivo functional flexion-extension and to evaluate the effects of anterior cervical arthrodesis on spine kinetics. Spine and head kinematics from 16 subjects (six arthrodesis patients and ten asymptomatic controls) were determined during functional flexion-extension using dynamic stereo X-ray and conventional reflective markers. Subject-specific inverse dynamics models, including three flexor muscles and four extensor muscles attached to the skull, estimated the force applied to C2. Total force applied to C2 was not significantly different between arthrodesis and control groups at any 10 deg increment of head flexion-extension (all p values ≥ 0.937). Forces applied to C2 were smallest in the neutral position, increased slowly with flexion, and increased rapidly with extension. Muscle moment arms changed significantly during flexion-extension, and were dependent upon the direction of head motion. The results suggest that in vitro protocols and finite element models that apply constant loads to C2 do not accurately represent in vivo cervical spine kinetics. [DOI: 10.1115/1.4023524]

Keywords: cervical kinematics, adjacent segment disease, musculoskeletal model, muscle moment arm

Introduction

The most common surgical treatment for cervical spine disorders is anterior arthrodesis [1], a surgery that involves replacing the degenerated intervertebral disk with a bone graft and attaching a rigid plate to the adjacent vertebrae to help stabilize the joint. It is believed arthrodesis may lead to increased motion in adjacent vertebrae, resulting in overload [2–6] that leads to degeneration [4,7–12] and potentially additional surgery [10,13–16]. In an attempt to eliminate or minimize adjacent segment overload, several motion-preserving disk replacement devices have been recently developed [17–20].

The effects of degeneration and surgery on cervical spine mechanics are commonly evaluated through *in vitro* testing [6,21–25] and finite element models derived from *in vitro* tests [26–30]. Due to the fact that cervical spine specimens will buckle under a compressive load of approximately 10 N [31], *in vitro* test protocols often apply only a moment to the specimen to simulate *in vivo* kinematics [28,32,33]. However, pure-moment loading methods do not replicate the physiological response and are poorly suited for evaluating certain surgical procedures and instrumentation [34]. More recently, “follower load” protocols have been developed to increase the compressive force applied to spine and avoid bucking [30,35,36]. However, it is not clear how well these protocols replicate *in vivo* dynamic, functional loading.

An estimate of cervical spine loading may be obtained through inverse dynamics models. Previous inverse dynamics models of the cervical spine have primarily focused on the response of the head and neck to impacts [37–40]. The biofidelity of these models may be increased by including muscular activation in the model [41]. However, it is difficult to accurately and reliably obtain muscular activation parameters for deep neck muscles during dynamic movement [42].

The objectives of the current study were to estimate the load applied to the C2 vertebra during *in vivo* functional flexion-extension and to evaluate the effects of anterior cervical arthrodesis on spine loads. The load applied to the superior cervical spine (C2) was estimated using subject-specific kinematic motion data obtained from conventional motion analysis (for head motion) and biplane radiography (for vertebral motion), in combination with muscular attachment and maximal muscle force parameters obtained from the literature. The first hypothesis tested was that spinal loads would not be significantly different in asymptomatic control subjects and single-level anterior arthrodesis patients. The second hypothesis tested was that the load applied to C2 would change significantly throughout the range of head flexion-extension.

Materials and Methods

Following Institutional Review Board (IRB) approval, data were analyzed from six single-level anterior arthrodesis patients (average age: 43.2 ± 8.8 years; 2 M, 4 F; 3 C56 arthrodesis, 3 C67 arthrodesis; 7.0 ± 0.6 months post-surgery) and ten asymptomatic control subjects of similar age (average age: 44.7 ± 6.6 years; 3 M, 7 F) who provided informed consent to participate in this

¹Corresponding author.

Contributed by the Bioengineering Division of ASME for publication in the JOURNAL OF BIOMECHANICAL ENGINEERING. Manuscript received October 5, 2012; final manuscript received January 7, 2013; accepted manuscript posted January 29, 2013; published online May 9, 2013. Assoc. Editor: Yener N. Yeni.

Fig. 1 The biplane X-ray system. X-ray tubes (left) directed X-rays through the subject to image intensifiers (right). 2.5 ms X-ray pulses (70 kV, 160 mA) were generated by cardiac cine-angiography generators at a rate of 30 Hz and radiographic images were collected by high-speed cameras synchronized to the X-ray pulses.

research study. High-resolution CT scans ($0.29 \times 0.29 \times 1.25$ mm voxels) of the cervical spine (C2–C7) were acquired on each participant (GE Lightspeed 16, GE Medical Systems, Waukesha, WI). Bone tissue was segmented from the CT volume using a combination of commercial software (Mimics software, Materialise, Leuven, Belgium) and manual segmentation [43]. A three-dimensional (3D) model of each vertebra was generated from the segmented bone tissue. Three-dimensional male and female skull models were generated using male and female data from the Visible Human dataset [44] and scaled for each subject according to subject-specific anterior to posterior skull dimensions obtained from the sagittal scout view prior to the CT scan.

Participants were seated within a biplane X-ray system (Fig. 1) and, for each trial, directed to continuously move their head and neck through their entire range of flexion-extension. A metronome set at 40 to 44 beats per min ensured the participants moved at a continuous, steady pace to complete each full movement cycle in 3 s or less. Biplane radiographs were collected simultaneously, at 30 frames per s for 3 s, as the subject performed continuous flexion-extension. Radiographs were recorded for two or three separate trials for each subject (allowing for a rest period between trials), resulting in a total of 42 movement trials analyzed for this study (25 control trials and 17 arthrodesis trials). A trial with the subject stationary, looking forward with the head in the neutral position, was also collected for each participant. Reflective markers were placed on the head and torso to determine head motion using conventional motion analysis techniques (sample rate 60 Hz; Vicon-MX, Oxford, UK).

A previously validated volumetric model-based tracking process determined three-dimensional vertebral position with sub-

millimeter accuracy [45] for all static and dynamic trials. Details describing the tracking process, including hardware and software specifications, calibration and distortion correction procedures, and computational algorithms have been described previously [45–48]. Three-dimensional bone positions and reflective marker data were filtered at 1.0 and 10.0 Hz, respectively, using a fourth-order, low-pass Butterworth filter with the filter frequency determined using residual analysis [49]. A calibration object containing radiopaque and reflective markers was used to co-register tracked vertebrae and head motion.

The skull orientation relative to the head reflective markers was determined from the static neutral trial by manually aligning the foramen magnum to the spinal canal. Correct alignment between the skull and C2 was verified using three-dimensional animations of the flexion-extension movement. For each trial, it was visually confirmed that the foramen magnum remained aligned with the spinal canal passing through C2 as the skull and cervical spine moved through the entire range of flexion-extension.

The musculoskeletal model included three flexor muscles that were active only when the head was in an extended orientation relative to the neutral position (sternocleidomastoid (SCM), longus capitis (LGC), and infrahyoid), and four extensor muscles that were active only when the head was in a flexed orientation relative to the neutral position (trapezius (Trap), splenius capitis (SPL), semispinalis capitis (SSP), and rectus capitis posterior major (RCP)). Attachment sites [50–52] and maximal force output [50,52] for each muscle were obtained from the literature (Table 1). The longus capitis muscle was required to wraparound C2 and C3 when the vector from its origin to insertion intersected these bones, while all other muscles were modeled using straight line paths. The infrahyoid attachment was defined as 35 mm anterior and 3 mm inferior to the C3 vertebra [53], and its location remained constant relative to C3 [54]. Muscle insertion sites were identified on the 3D skull models (and therefore did not vary among males nor among females, although they were scaled along with the size of the skull), while muscle origin sites were determined from subject-specific reflective marker data (in the case of trapezius, infrahyoid, and sternocleidomastoid) and subject-specific vertebral anatomy (for the remaining four muscles) (Fig. 2). Head mass (male = 4.2 kg; female = 3.4 kg), center of mass (59 mm superior and 18 mm anterior to the occipital condyle) and moment of inertia about the skull anatomical axes values were obtained from the literature [32].

Two assumptions were made to determine the force applied by the skull on C2. The first was to require equal activation in all active muscles (i.e., the same percentage of maximum muscle force). Second, zero cocontraction was assumed in antagonistic muscles. With these assumptions in place, the force applied by the skull on C2 was determined by solving the equations of dynamic equilibrium as follows: The midpoint of the superior facets on C2 was used to represent the single point on C2 where the head applied force to the cervical spine. The muscle moment arms for each frame of data were determined for each muscle given the head, torso, and spine position obtained from the tracked reflective and radiographic data. This was accomplished by calculating the

Table 1 Muscle attachments and peak forces [50–52]

Muscle	Origin	Insertion	Peak force (N)
Sternocleidomastoid	Sternum	Mastoid process	344
Longus Capitis	C4 anterior tubercle	Basilar part of occipital bone	96
Infrahyoid (omohyoid, sternohyoid, sternothyroid)	Sternum	Hyoid bone	138
Trapezius	Clavicle	Superior nuchal line of occipital bone	264
Splenius capitis	C7 spinous process	Mastoid process of temporal bone/superior nuchal line of occipital bone	216
Semispinalis capitis	C6 articular process	Between superior and inferior nuchal line of occipital bone	386
Rectus capitis posterior major	C2 spinous process	Anterior to inferior nuchal line of occipital bone	181

Fig. 2 The skull and cervical spine model for a representative subject in an (a) extended orientation, (b) the neutral position, and (c) a flexed orientation. Reflective markers (green) determined skull and torso position, while vertebral positions were determined from model-based tracking of biplane radiographs. Muscle forces (red vectors) were generated by flexor muscles (yellow spheres identify insertions) when the head was in an extended orientation and by extensor muscles (magenta spheres identify insertions) when the head was in a flexed orientation. The orange vector represents the weight of head. The teal vector represents the total force applied by the head to C2. Anatomic coordinate systems were generated within each vertebra (only C7 is shown for clarity).

perpendicular distance from the midpoint of the C2 superior facets to the line of action of the muscle. The maximum torque that could be generated ($T_{\max}$) was then determined using the muscle moment arms and corresponding maximal muscle forces (Table 1). Next, the torque required by the muscles to counter the torque generated by the skull angular acceleration and the weight of the head was determined: $\sum T_m = I_{\text{skull}_{C2PT}} \times \alpha_{\text{skull}} - T_{WH}$, where T_m is the total muscle torque, $I_{\text{skull}_{C2PT}}$ is the moment of inertia of the skull about the midpoint of the superior facets on C2, α_{skull} is the angular acceleration of the skull (obtained from head reflective markers) and T_{WH} is the torque due to the weight of the head. The torque required by the muscles was expressed as a percentage of maximum torque: $T_{\text{Percent}} = \sum T_m / T_{\max}$. The magnitude of each muscle force was then calculated by multiplying each maximum muscle force by T_{Percent} . The components of each muscle force were calculated from the muscle magnitudes and lines of action (from muscle attachments). Then the total force applied to C2 was calculated: $F_{C2} = m_H a_H - F_{WH} - \sum F_m$, where F_{C2} was the force applied to C2, $m_H a_H$ was the mass and acceleration of the head, F_{WH} was the force due to the weight of the head, and $\sum F_m$ was the sum of the muscle forces. Two-dimensional motion was assumed and muscular attachment sites and activation levels were assumed to be symmetric about the sagittal plane.

Cervical spine kinetics were characterized by the following parameters: the total force applied to C2 due to muscular and inertial forces, and the moment arm of each muscle throughout the range of flexion-extension.

Head flexion-extension angle, normalized to the static neutral trial, was calculated for each tracked data frame and interpolated at 1 deg intervals for each trial. Forces and moment arm values that were calculated at each tracked radiographic frame were then interpolated to correspond to each 1 deg increment in head flexion-extension angle. Multiple trials from the same subject were averaged to yield a single average dataset for each subject used for statistical analysis.

Statistical analysis was carried out using data corresponding to 10 deg increments in head flexion-extension, from 40 deg of extension (−40 deg) to 40 deg of flexion. An independent-samples *t* test was used to test for differences between asymptomatic and arthrodesis groups in the total force applied to C2. A paired-samples *t* test was used to identify differences in the force applied to C2 at identical angles of head flexion-extension during the flexion and extension movement directions. Repeated measures analysis of variance was used to identify significant differences in muscle moment arms at each 10 deg increment of head

flexion-extension. Significance was set at $p < 0.05$ for all tests. *P* values were adjusted for multiple comparisons using the false discovery rate [55].

The sensitivity of the model to muscle attachment sites was evaluated by adjusting the SCM and SSP attachments to the skull and the infrahyoid attachment to the hyoid by ± 8 , ± 4 , ± 2 , and ± 1 mm in the anterior (+) and posterior (−) directions of the skull coordinate system. The model was run with each of these independently altered attachment sites for the SCM, SSP, and infrahyoid. The percent change in total force applied to C2 was calculated for each altered attachment site at 10 deg increments of head flexion-extension.

Results

Total Force Applied to C2. The total force applied to C2 was not different between asymptomatic and arthrodesis groups at any 10 deg interval of head flexion-extension, from 40 deg of extension to 40 deg of flexion (average absolute difference: 8.2 N; range of differences between groups: −32 N to +21 N; all *p* values ≥ 0.937). Therefore, data from asymptomatic and arthrodesis subjects were combined to determine if the force applied to C2 differed during the flexion and extension movements.

The total force applied to C2 followed a consistent pattern within subjects. In the flexed orientation (relative to the static neutral position), the force applied to C2 was slightly higher when the head was moving in the direction of flexion than in the direction of extension. When the head was in an extended orientation, the force applied to C2 was higher when the head was moving in the direction of extension. These differences in total force applied to C2 during the flexion and extension motions were statistically significant at all head orientations except for the neutral position (Fig. 3). The average within-subject trial-to-trial variability (i.e., standard deviation) in the total force applied to C2 averaged 8.6% of the total force applied to C2 over the range of motion from 40 deg of extension to 40 deg of flexion.

Flexor Muscle Moment Arms. Flexor muscle moment arms were not significantly different between asymptomatic and arthrodesis groups at any 10 deg increment of head extension from neutral to 40 deg extension (SCM: average absolute difference between groups: 1.3 mm, range of differences between groups: 0.2 to 2.2 mm, all $p \geq 0.850$; LGC: average absolute difference between groups: 0.6 mm, range of differences between groups:

Fig. 3 The total force made by the skull on C2 while the head moved in flexion (blue diamonds) and extension (red squares). Data points indicate the average of all subjects (asymptomatic plus arthrodesis). Error bars indicate intersubject variability (the 95% confidence interval of the mean), however, within-subject comparisons were performed to identify differences between flexion and extension movement directions. *P* values comparing flexion versus extension, adjusted for multiple comparisons, are provided for each 10 deg increment of head flexion-extension. The horizontal axis indicates head flexion-extension angle relative to the neutral position.

Fig. 4 The perpendicular distance from the midpoint of the C2 superior facets to the line of action of the flexor muscles during the flexion (solid lines) and extension (dashed lines) motion. Data points indicate the average of all subjects (asymptomatic plus arthrodesis). Positive values indicate a flexion moment, negative values indicate an extension moment. Error bars indicate intersubject variability (the 95% confidence interval of the mean), however, within-subject comparisons were performed to identify differences between flexion and extension movement directions and to identify differences among 10 deg increments of head movement. Each flexion value was significantly greater than the corresponding extension value for longus capitus (LGC) (all $p \leq 0.003$) and infrahyoid (all $p \leq 0.002$) but only at 0 deg, 10 deg, and 20 deg of extension for the SCM moment arm ($p \leq 0.033$). For the SCM and LGC, the average moment arm (flexion + extension movement direction) significantly decreased at each 10 deg increment of head extension from neutral to 40 deg of extension (all $p \leq 0.020$ and all $p \leq 0.001$, respectively). The infrahyoid moment arm significantly increased each 10 deg of head flexion-extension from neutral to 40 deg of extension (all $p \leq 0.001$).

−0.5 to 0.9 mm, all $p \geq 0.828$; infrahyoid: average absolute difference between groups: 5.9 mm, range of differences between groups: 2.9 to 9.5 mm, all $p \geq 0.197$). Therefore, asymptomatic and arthrodesis subjects were grouped for further statistical analysis.

Flexor muscle moment arms were consistently greater when the head was moving in flexion versus extension (Fig. 4). These differences in muscle moment arms when moving in flexion versus extension were significant at each 10 deg increment of head motion for the LGC and infrahyoid muscles (all $p \leq 0.003$ and all

$p \leq 0.002$, respectively), but only at 0 deg, 10 deg, and 20 deg of extension for the SCM moment arm ($p \leq 0.033$) (Fig. 4).

The SCM moment arm changed significantly each 10 deg of head flexion-extension from neutral to 40 deg of extension (all $p \leq 0.020$). On average, the sternocleidomastoid muscle functioned as a head flexor when the head was within 25 deg of the neutral position, and as the head orientation progressed beyond 25 deg of extension, the sternocleidomastoid muscle began to function as a head extensor. The LGC moment arm significantly decreased each 10 deg of head flexion-extension from neutral to 40 deg of extension (all $p \leq 0.001$), while the infrahyoid moment arm significantly increased each 10 deg of head flexion-extension from neutral to 40 deg of extension (all $p \leq 0.001$) (Fig. 4).

Extensor Muscle Moment Arms. Extensor muscle moment arms were not significantly different between asymptomatic and arthrodesis groups at any 10 deg increment of head flexion from neutral to 40 deg flexion (Trap: average absolute difference between groups: 1.8 mm, range of differences between groups: 0.8 to 3.5 mm, all $p \geq 0.832$; SPL: average absolute difference between groups: 3.2 mm, range of differences between groups: -4.2 to -2.2 mm, all $p \geq 0.373$; SSP: average absolute difference between groups: 0.5 mm, range of differences between groups: -0.9 to 0.8 mm, all $p \geq 0.875$; RCP: average absolute difference between groups: 0.4 mm, range of differences between groups: -1.1 to 0.0 mm, all $p \geq 0.961$). Therefore, asymptomatic and arthrodesis subjects were grouped for further statistical analysis.

Trapezius, splenius capitis, and semispinalis muscle moment arms significantly increased at each successive 10 deg increment of head motion when the head was moving in extension (all $p \leq 0.001$ for trapezius and semispinalis, all $p \leq 0.002$ for splenius capitis) (Fig. 5). The rectus capitis posterior muscle moment arm

was significantly greater when the head was moving in extension from 10 deg to 40 deg of head flexion (all $p \leq 0.022$).

Trapezius and splenius moment arms decreased significantly each 10 deg of head flexion from neutral to 40 deg of flexion (all $p \leq 0.016$ and all $p \leq 0.038$, respectively). Similarly, the semispinalis moment arm decreased significantly each 10 deg of head flexion from neutral to 30 deg of flexion (all $p \leq 0.015$). The rectus capitis posterior major moment arm significantly decreased from 30 deg to 40 deg of head flexion ($p = 0.019$) (Fig. 5).

Sensitivity Analysis. The model was relatively insensitive to changes in the semispinalis attachment location. The maximum change in total force applied to C2 was less than 5%, and occurred at 40 deg of flexion with a +8 mm adjustment to the semispinalis attachment site. The model was more sensitive to changes in the infrahyoid attachment site, estimating up to a 24% increase in the total force applied to C2 when the head was at 40 deg of extension. The model was most sensitive to changes in the sternocleidomastoid attachment site, estimating a 59% increase in the total force applied to C2 with a -8 mm adjustment to the sternocleidomastoid attachment site and the head at 40 deg of extension (Fig. 6).

Discussion

The current *in vivo* data indicate that inertial and muscular forces combine to exert force on the upper cervical spine in a predictable pattern related to head orientation during flexion-extension. The model estimated the force applied by the head to C2 is smallest in the neutral position and increased slowly with head flexion and increased rapidly with head extension. The statistically significant differences in force applied to C2 at corresponding head angles when moving in flexion and in extension have not been reported or discussed previously. The differences in force

Fig. 5 The perpendicular distance from the midpoint of the C2 superior facets to the line of action of the extensor muscles during the flexion (solid lines) and extension (dashed lines) motion. Data points indicate the average of all subjects (asymptomatic plus arthrodesis). Negative values indicate an extension moment. Error bars indicate intersubject variability (the 95% confidence interval of the mean), however, within-subject comparisons were performed to identify differences between flexion and extension movement directions and to identify differences among 10 deg increments of head extension. Each extension value was significantly greater than the corresponding flexion value for the trapezius (Trap) (all $p \leq 0.001$), semispinalis (SSP) (all $p \leq 0.001$), and splenius capitis (SPL) (all $p \leq 0.002$). Each extension value was significantly greater than the corresponding flexion value for the rectus capitis posterior major (RCP) from 10 deg to 40 deg of head flexion (all $p \leq 0.022$). The Trap and SPL average moment arms (flexion + extension movement direction) were significantly different at each 10 deg increment of head flexion (all $p \leq 0.016$ and all $p \leq 0.038$, respectively), while the SSP average moment arm decreased significantly at each 10 deg increment of head flexion from neutral to 30 deg flexion (all $p \leq 0.015$). The rectus capitis posterior major moment arm significantly decreased from 30 deg to 40 deg of head flexion ($p = 0.019$).

Sensitivity to SCM Muscle Attachment Site

Fig. 6 Sensitivity of the model to changes in the sternocleidomastoid attachment site on the skull. The vertical axis is the percent change in force applied to C2, relative to the force determined using the original attachment site. The attachment site was repositioned 8, 4, 2, or 1 mm anterior (+) or posterior (–) to the original attachment site. Changing the SSP and infrahyoid attachment sites yielded changes in the C2 force curves that were smaller in magnitude but similar in pattern.

stem from differences in muscle moment arms in flexion and extension for identical head angles (Figs. 4 and 5). As we have reported previously, during flexion-extension, the head leads the motion and the vertebrae follow the head [56]. Therefore, the spine orientation relative to the head is not identical when moving in flexion and in extension for any given head angle. From this, it follows that the muscle moment arms are different when moving in flexion and in extension for any given head angle.

The present model suggests the force applied to the top of C2 reaches maximum values of approximately 200 N (or about 5 times head weight) at 40 deg of head extension and 100 N (or about 2.5 times head weight) at 40 deg of head flexion. In comparison, the literature contains wide-ranging estimates of cervical spine loads. It has been suggested that the compressive force applied to the cervical spine approaches three times the weight of the head (approximately 120 N) due to muscle coactivation forces balancing the head in the neutral, relaxed posture, and that minimal to moderate isometric muscle effort increases this compressive force to 250 N, however, no experimental or modeling data were provided to support this assertion [35]. A previous computational model predicted cervical spine forces from isometric contraction data and suggested that the compressive load is estimated to reach 1200 N in activities involving maximal isometric muscle efforts [41]. In agreement with the current results, this previous model estimated the compressive force at C2 in the neutral position, due to forces from muscles attached to the skull and head weight totaled approximately 53 N [41]. The current results are also in agreement with Hattori who measured *in vivo* disk pressure in various head orientations and found the highest disk pressure in extension (approximately equivalent to a compression force of 155 N) and lowest disk pressure in the neutral position (approximately equivalent to a compression force of 63 N) [30,35,57].

The change in the force applied to C2 that occurred with head flexion-extension and the differences in force applied to C2 with movement direction suggest that cervical spine testing protocols that apply a constant follower load to the specimen may not accurately represent *in vivo* kinetics. It is possible the biofidelity of *in vitro* tests may be improved by reproducing the shape of the force curves presented in Fig. 3.

The current study is believed to be the first to estimate changes in muscle moment arm during active *in vivo* flexion-extension. The results indicate that muscle moment arms change significantly

during flexion-extension, and the magnitude of muscle moment arms are significantly different during flexion and extension movements. These results suggest computational models of spine loading should account for head orientation and direction of movement when determining moment arms for muscles attached to the skull. This *in vivo* data confirms a previous computational model that concluded that variations in muscle moment arms throughout the range of motion can alter muscle moment-generating capacities [58]. The moment arms of the muscles attached to the hyoid appear to be the most dependent on head flexion angle. The importance of including muscles attached to the hyoid in cervical spine models has been noted previously [42,50,52]. Results from the current study suggest the infrahyoid muscles may be particularly important when the head approaches full extension, as this is the orientation in which infrahyoid moment arm is largest (thus permitting the generation of a large flexor moment) and simultaneously the sternocleidomastoid muscles are exerting an extensor moment about the upper cervical spine. This interpretation of the results suggests that when the head is near full extension, the required flexion torque may be generated, and forces on the cervical spine minimized, if sternocleidomastoid activation is decreased and infrahyoid activation is increased. There are currently no *in vivo* data to suggest infrahyoid muscles increase their activation and sternocleidomastoid muscles decrease activation in order to minimize forces on the spine at large head extension angles.

The muscle moment arms determined in the current study, calculated geometrically (perpendicular distance method), are in general agreement with a recent cadaver-based study of neck muscle moment arms at C0–C1 measured by the tendon excursion method (SCM: 0.1 cm; Trap: 4.6 cm; SPL: 1.3 cm; SSP: 4.9 cm; RCP: 1.9 cm) [51]. A previous computational model of the neck included some muscle moment arms in the neutral position that were in agreement with values determined in the current study (SCM: 1 cm; LCG: 1 cm; Trap 5 cm) and some muscle moment arms that were substantially larger than in the current study (SPL: 5 cm; SSP: 6.2 cm) [58]. The current study, in agreement with previous cadaveric studies [51,52], indicates that the sternocleidomastoid acts as a head flexor and extensor.

One simplifying assumption of the model was that each muscle attachment site could be represented by a single point. The model sensitivity to the location of this point varied by muscle (low sensitivity to SSP attachment site, high sensitivity to SCM attachment

site), head flexion-extension angle (sensitivity increased as the head moved from the neutral position), and distance from the original attachment site (the model was relatively insensitive to moving the muscle attachment 4 mm or less from the original position, but highly sensitive to an 8 mm change in muscle attachment site). These results suggest that if subject-specific attachment sites are included in the model (perhaps from MRI images), the model results will be substantially affected by errors in muscle attachment site in excess of ± 4 mm. The sensitivity analysis of the hyoid and sternocleidomastoid are particularly informative, as they indicate model sensitivity to hyoid kinematics (the hyoid was assumed fixed relative to C3) and model sensitivity to the broad attachment between the sternocleidomastoid and the skull.

As with any computational model, there are assumptions and limitations associated with the model and results. The muscles that were selected for inclusion in the model have been established as the major force generating muscles responsible for flexion-extension of the skull [42,50,52]. However, the muscle models do not include characteristics such as the force-length-velocity relationship, active and passive stiffness parameters, muscle fiber length, and pennation angle. While these muscle parameters could be included in future model development, it is not clear that a more complex model would yield additional functional or clinically significant information. The assumption that cocontraction is minimal in the neck muscles has been supported by EMG data [42] and this assumption has been made previously [41]. It is extremely difficult to acquire muscle activation patterns (let alone muscle force estimates) from deep muscles of the neck. Therefore, there are no reliable data available to provide a basis for the magnitude and timing of muscle cocontraction during dynamic, functional movement. Previous fine-wire EMG data for the anterior neck muscles [42] support the common modeling assumption of uniform activation used in the current model. The relatively small sample size limited the power of the study to detect significant differences between groups if they did, in fact, exist. An additional limitation is that the current study only estimated forces applied to the C2 vertebra. Forces applied to inferior vertebrae are affected by muscles that attach these vertebrae to more inferior bones. More accurate estimates of the effects of arthrodesis on tissue-level cervical disk mechanics may be determined by increasing the complexity of the model by adding muscles that attach the vertebrae to the torso. Additionally, subject-specific finite element models that are subjected to loads that change in magnitude as the head moves through flexion-extension may reveal important tissue-level differences between asymptomatic and arthrodesis patients.

Acknowledgment

Sources of support: NIH/NIAMS Grant No. 1R03AR056265 and a 21st Century Developmental Grant from The Cervical Spine Research Society.

References

- Patil, P. G., Turner, D. A., and Pietrobon, R., 2005, "National Trends in Surgical Procedures for Degenerative Cervical Spine Disease: 1990–2000," *Neurosurgery*, **57**(4), pp. 753–758.
- Fielding, J. W., 1964, "Normal and Selected Abnormal Motion of the Cervical Spine From the Second Cervical Vertebra to the Seventh Cervical Vertebra Based on Cineroentgenography," *J. Bone Joint Surg.*, **46**, pp. 1779–1781.
- Dunsker, S. B., Colley, D. P., and Mayfield, F. H., 1978, "Kinematics of the Cervical Spine," *Clin. Neurosurg.*, **25**, pp. 174–183.
- Baba, H., Furusawa, N., Imura, S., Kawahara, N., Tsuchiya, H., and Tomita, K., 1993, "Late Radiographic Findings After Anterior Cervical Fusion for Spondylotic Myeloradiculopathy," *Spine*, **18**(15), pp. 2167–2173.
- Matsunaga, S., Kabayama, S., Yamamoto, T., Yone, K., Sakou, T., and Nakanishi, K., 1999, "Strain on Intervertebral Discs After Anterior Cervical Decompression and Fusion," *Spine*, **24**(7), pp. 670–675.
- Schwab, J. S., Diangelo, D. J., and Foley, K. T., 2006, "Motion Compensation Associated With Single-Level Cervical Fusion: Where Does the Lost Motion Go?," *Spine*, **31**(21), pp. 2439–2448.
- Hunter, L. Y., Braunstein, E. M., and Bailey, R. W., 1980, "Radiographic Changes Following Anterior Cervical Fusion," *Spine*, **5**(5), pp. 399–401.
- Gore, D. R., and Sepic, S. B., 1998, "Anterior Discectomy and Fusion for Painful Cervical Disc Disease. A Report of 50 Patients With an Average Follow-Up of 21 Years," *Spine*, **23**(19), pp. 2047–2051.
- Hilibrand, A. S., Carlson, G. D., Palumbo, M. A., Jones, P. K., and Bohlman, H. H., 1999, "Radiculopathy and Myelopathy at Segments Adjacent to the Site of a Previous Anterior Cervical Arthrodesis," *J. Bone Joint Surg.*, **81**(4), pp. 519–528.
- Goffin, J., Geusens, E., Vantomme, N., Quintens, E., Waerzeggers, Y., Depreitere, B., Van Calenbergh, F., and van Loon, J., 2004, "Long-Term Follow-Up After Interbody Fusion of the Cervical Spine," *J. Spinal Disord. Tech.*, **17**(2), pp. 79–85.
- Ishihara, H., Kanamori, M., Kawaguchi, Y., Nakamura, H., and Kimura, T., 2004, "Adjacent Segment Disease After Anterior Cervical Interbody Fusion," *Spine J.*, **4**(6), pp. 624–628.
- Kulkarni, V., Rajshekhar, V., and Raghuram, L., 2004, "Accelerated Spondylotic Changes Adjacent to the Fused Segment Following Central Cervical Corpectomy: Magnetic Resonance Imaging Study Evidence," *J. Neurosurg.*, **100**(1 Suppl. Spine), pp. 2–6.
- Bohlman, H. H., Emery, S. E., Goodfellow, D. B., and Jones, P. K., 1993, "Robinson Anterior Cervical Discectomy and Arthrodesis for Cervical Radiculopathy. Long-Term Follow-Up of One Hundred and Twenty-Two Patients," *J. Bone Joint Surg.*, **75**(9), pp. 1298–1307.
- Gore, D. R., and Sepic, S. B., 1984, "Anterior Cervical Fusion for Degenerated or Protruded Discs. A Review of One Hundred Forty-Six Patients," *Spine*, **9**(7), pp. 667–671.
- Watters, W. C., 3rd, and Levinthal, R., 1994, "Anterior Cervical Discectomy With and Without Fusion. Results, Complications, and Long-Term Follow-Up," *Spine*, **19**(20), pp. 2343–2347.
- Hilibrand, A. S., Yoo, J. U., Carlson, G. D., and Bohlman, H. H., 1997, "The Success of Anterior Cervical Arthrodesis Adjacent to a Previous Fusion," *Spine*, **22**(14), pp. 1574–1579.
- Sasso, R. C., Anderson, P. A., Riew, K. D., and Heller, J. G., 2011, "Results of Cervical Arthroplasty Compared With Anterior Discectomy and Fusion: Four-Year Clinical Outcomes in a Prospective, Randomized Controlled Trial," *J. Bone Joint Surg.*, **93**(18), pp. 1684–1692.
- Jawahar, A., Cavanaugh, D. A., Kerr, E. J., 3rd, Birdsong, E. M., and Nunley, P. D., 2010, "Total Disc Arthroplasty Does Not Affect the Incidence of Adjacent Segment Degeneration in Cervical Spine: Results of 93 Patients in Three Prospective Randomized Clinical Trials," *Spine J.*, **10**(12), pp. 1043–1048.
- McAfee, P. C., Reah, C., Gilder, K., Eisermann, L., and Cunningham, B., 2011, "A Meta-Analysis of Comparative Outcomes Following Cervical Arthroplasty or Anterior Cervical Fusion: Results From Four Prospective Multi-Center Randomized Clinical Trials and Up to 1226 Patients," *Spine*, May 15, **37**(11), pp. 943–952.
- Kelly, M. P., Mok, J. M., Frisch, R. F., and Tay, B. K., 2011, "Adjacent Segment Motion After Anterior Cervical Discectomy and Fusion Versus Prodisc-C Cervical Total Disk Arthroplasty: Analysis From a Randomized, Controlled Trial," *Spine*, **36**(15), pp. 1171–1179.
- DiAngelo, D. J., Roberston, J. T., Metcalf, N. H., McVay, B. J., and Davis, R. C., 2003, "Biomechanical Testing of an Artificial Cervical Joint and an Anterior Cervical Plate," *J. Spinal Disord. Tech.*, **16**(4), pp. 314–323.
- Brodke, D. S., Klimo, P., Jr., Bachus, K. N., Braun, J. T., and Dailey, A. T., 2006, "Anterior Cervical Fixation: Analysis of Load-Sharing and Stability With Use of Static and Dynamic Plates," *J. Bone Joint Surg.*, **88**(7), pp. 1566–1573.
- Davies, M. A., Bryant, S. C., Larsen, S. P., Murrey, D. B., Nussman, D. S., Laxer, E. B., and Darden, B. V., 2006, "Comparison of Cervical Disk Implants and Cervical Disk Fusion Treatments in Human Cadaveric Models," *J. Biomech. Eng.*, **128**(4), pp. 481–486.
- Dmitriev, A. E., Cunningham, B. W., Hu, N., Sell, G., Vigna, F., and McAfee, P. C., 2005, "Adjacent Level Intradiscal Pressure and Segmental Kinematics Following a Cervical Total Disc Arthroplasty: An *In Vitro* Human Cadaveric Model," *Spine*, **30**(10), pp. 1165–1172.
- Eck, J. C., Humphreys, S. C., Lim, T. H., Jeong, S. T., Kim, J. G., Hodges, S. D., and An, H. S., 2002, "Biomechanical Study on the Effect of Cervical Spine Fusion on Adjacent-Level Intradiscal Pressure and Segmental Motion," *Spine*, **27**(22), pp. 2431–2434.
- Brolin, K., and Halldin, P., 2004, "Development of a Finite Element Model of the Upper Cervical Spine and a Parameter Study of Ligament Characteristics," *Spine*, **29**(4), pp. 376–385.
- del Palomar, A. P., Calvo, B., and Doblare, M., 2008, "An Accurate Finite Element Model of the Cervical Spine Under Quasi-Static Loading," *J. Biomech.*, **41**(3), pp. 523–531.
- Kallemeyn, N., Gandhi, A., Kode, S., Shivanna, K., Smucker, J., and Grosland, N., 2010, "Validation of a C2–C7 Cervical Spine Finite Element Model Using Specimen-Specific Flexibility Data," *Med. Eng. Phys.*, **32**(5), pp. 482–489.
- Maiman, D. J., Kumaresan, S., Yoganandan, N., and Pintar, F. A., 1999, "Biomechanical Effect of Anterior Cervical Spine Fusion on Adjacent Segments," *Biomed. Mater. Eng.*, **9**(1), pp. 27–38.
- Womack, W., Leahy, P. D., Patel, V. V., and Puttlitz, C. M., 2011, "Finite Element Modeling of Kinematic and Load Transmission Alterations Due to Cervical Intervertebral Disc Replacement," *Spine*, Aug 1, **36**(17), pp. E1126–E1133.
- Panjabi, M. M., Cholewicki, J., Nibu, K., Grauer, J., Babat, L. B., and Dvorak, J., 1998, "Critical Load of the Human Cervical Spine: An *In Vitro* Experimental Study," *Clin. Biomechan.*, **13**(1), pp. 11–17.
- Yoganandan, N., Pintar, F. A., Zhang, J., and Baisden, J. L., 2009, "Physical Properties of the Human Head: Mass, Center of Gravity and Moment of Inertia," *J. Biomech.*, **42**(9), pp. 1177–1192.

- [33] Panjabi, M. M., Crisco, J. J., Vasavada, A., Oda, T., Cholewicki, J., Nibu, K., and Shin, E., 2001, "Mechanical Properties of the Human Cervical Spine As Shown by Three-Dimensional Load-Displacement Curves," *Spine*, **26**(24), pp. 2692–2700.
- [34] DiAngelo, D. J., and Foley, K. T., 2004, "An Improved Biomechanical Testing Protocol for Evaluating Spinal Arthroplasty and Motion Preservation Devices in a Multilevel Human Cadaveric Cervical Model," *Neurosurg. Focus*, **17**(3), p. E4.
- [35] Patwardhan, A. G., Havey, R. M., Ghanayem, A. J., Diener, H., Meade, K. P., Dunlap, B., and Hodges, S. D., 2000, "Load-Carrying Capacity of the Human Cervical Spine in Compression is Increased Under a Follower Load," *Spine*, **25**(12), pp. 1548–1554.
- [36] Patwardhan, A. G., Havey, R. M., Meade, K. P., Lee, B., and Dunlap, B., 1999, "A Follower Load Increases the Load-Carrying Capacity of the Lumbar Spine in Compression," *Spine*, **24**(10), pp. 1003–1009.
- [37] Seacrist, T., Arbogast, K. B., Maltese, M. R., Garcia-Espana, J. F., Lopez-Valdes, F. J., Kent, R. W., Tanji, H., Higuchi, K., and Balasubramanian, S., 2012, "Kinetics of the Cervical Spine in Pediatric and Adult Volunteers During Low Speed Frontal Impacts," *J. Biomech.*, **45**(1), pp. 99–106.
- [38] Pintar, F. A., Yoganandan, N., and Baisden, J., 2005, "Characterizing Occipital Condyle Loads Under High-Speed Head Rotation," *Stapp Car Crash J.*, **49**, pp. 33–47.
- [39] Pintar, F. A., Yoganandan, N., and Maiman, D. J., 2010, "Lower Cervical Spine Loading in Frontal Sled Tests Using Inverse Dynamics: Potential Applications for Lower Neck Injury Criteria," *Stapp Car Crash J.*, **54**, pp. 133–166.
- [40] Ivancic, P. C., Panjabi, M. M., and Ito, S., 2006, "Cervical Spine Loads and Intervertebral Motions During Whiplash," *Traffic Inj. Prev.*, **7**(4), pp. 389–399.
- [41] Moroney, S. P., Schultz, A. B., and Miller, J. A., 1988, "Analysis and Measurement of Neck Loads," *J. Orthop. Res.*, **6**(5), pp. 713–720.
- [42] Siegmund, G. P., Blouin, J. S., Brault, J. R., Hedenstierna, S., and Inglis, J. T., 2007, "Electromyography of Superficial and Deep Neck Muscles During Isometric, Voluntary, and Reflex Contractions," *J. Biomech. Eng.*, **129**(1), pp. 66–77.
- [43] Thorhauer, E., Miyawaki, M., Illingworth, K., Holmes, A., and Anderst, W., 2010, "Accuracy of Bone and Cartilage Models Obtained From CT and MRI," American Society of Biomechanics, Providence, RI.
- [44] University of Iowa, downloaded 03/29/2012, https://mri.radiology.uiowa.edu/visible_human_datasets.html
- [45] Anderst, W. J., Baillargeon, E., Donaldson, W. F., 3rd, Lee, J. Y., and Kang, J. D., 2011, "Validation of a Noninvasive Technique to Precisely Measure *In Vivo* Three-Dimensional Cervical Spine Movement," *Spine*, **36**(6), pp. E393–E400.
- [46] Bey, M. J., Zael, R., Brock, S. K., and Tashman, S., 2006, "Validation of a New Model-Based Tracking Technique for Measuring Three-Dimensional, *In Vivo* Glenohumeral Joint Kinematics," *J. Biomech. Eng.*, **128**(4), pp. 604–609.
- [47] Martin, D. E., Greco, N. J., Klatt, B. A., Wright, V. J., Anderst, W. J., and Tashman, S., 2011, "Model-Based Tracking of the Hip: Implications for Novel Analyses of Hip Pathology," *J. Arthroplasty*, **26**(1), pp. 88–97.
- [48] Anderst, W., Zael, R., Bishop, J., Dumps, E., and Tashman, S., 2009, "Validation of Three-Dimensional Model-Based Tibio-Femoral Tracking During Running," *Med. Eng. Phys.*, **31**(1), pp. 10–16.
- [49] Winter, D. A., 2009, *Biomechanics and Motor Control of Human Movement*, 4th ed., Wiley, Hoboken, NJ.
- [50] Chancey, V. C., Nightingale, R. W., Van Ee, C. A., Knaub, K. E., and Myers, B. S., 2003, "Improved Estimation of Human Neck Tensile Tolerance: Reducing the Range of Reported Tolerance Using Anthropometrically Correct Muscles and Optimized Physiologic Initial Conditions," *Stapp Car Crash J.*, **47**, pp. 135–153.
- [51] Ackland, D. C., Merritt, J. S., and Pandey, M. G., 2011, "Moment Arms of the Human Neck Muscles in Flexion, Bending and Rotation," *J. Biomech.*, **44**(3), pp. 475–486.
- [52] Oi, N., Pandey, M. G., Myers, B. S., Nightingale, R. W., and Chancey, V. C., 2004, "Variation of Neck Muscle Strength Along the Human Cervical Spine," *Stapp Car Crash J.*, **48**, pp. 397–417.
- [53] Andrade, A. V., Gomes, P. F., and Teixeira-Salmela, L. F., 2007, "Cervical Spine Alignment and Hyoid Bone Positioning With Temporomandibular Disorders," *J. Oral Rehabil.*, **34**(10), pp. 767–772.
- [54] Zheng, L., Jahn, J., and Vasavada, A. N., 2012, "Sagittal Plane Kinematics of the Adult Hyoid Bone," *J. Biomech.*, **45**(3), pp. 531–536.
- [55] Benjamini, Y., and Hochberg, Y., 1995, "Controlling the False Discovery Rate: A Practical and Powerful Approach to Multiple Testing," *J. R. Stat. Soc. Ser. B*, **57**(1), pp. 289–300.
- [56] Anderst, W. J., Donaldson, W. F., Lee, J. Y., and Kang, J. D., 2013, "Cervical Spine Intervertebral Kinematics With Respect to the Head are Different During Flexion and Extension Motions," *J. Biomech.*, (in press).
- [57] Hattori, S., Oda, H., and Kawai, U., 1981, "Cervical Intradiscal Pressure in Movements and Traction of the Cervical Spine," *Z. Orthop.*, **119**, pp. 568–569.
- [58] Vasavada, A. N., Li, S., and Delp, S. L., 1998, "Influence of Muscle Morphometry and Moment Arms on the Moment-Generating Capacity of Human Neck Muscles," *Spine*, **23**(4), pp. 412–422.