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INTER-INDIVIDUAL VARIATION IN VERTEBRAL KINEMATICS AFFECTS PREDICTIONS OF NECK MUSCULOSKELETAL MODELS

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Abstract

Experimental studies have found significant variation in cervical intervertebral kinematics (IVK) among healthy subjects, but the effect of this variation on biomechanical properties, such as neck strength, has not been explored. The goal of this study was to quantify variation in model predictions of extension strength, flexion strength and gravitational demand (the ratio of gravitational load from the weight of the head to neck muscle extension strength), due to inter-subject variation in IVK. IVK were measured from sagittal radiographs of twenty-four subjects (14F, 10M) in five postures: maximal extension, mid-extension, neutral, mid-flexion, and maximal flexion. IVK were defined by the position (anterior-posterior and superior-inferior) of each cervical vertebra with respect to T1 and its angle with respect to horizontal, and fit with a cubic polynomial over the range of motion. The IVK of each subject were scaled and incorporated into musculoskeletal models to create models that were identical in muscle force- and moment-generating properties but had subject-specific kinematics. The effect of inter-subject variation in IVK was quantified using the coefficient of variation (COV), the ratio of the standard deviation to the mean. COV of extension strength ranged from 8 – 15% over the range of motion, but COV of flexion strength were 20 – 80%. Moreover, the COV of gravitational demand was 80 – 90%, because the gravitational demand is affected by head position as well as neck strength. These results indicate that including inter-individual variation in models is important for evaluating neck musculoskeletal biomechanical properties.

Keywords

neck models; cervical spine kinematics; neck strength; ergonomics

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CONFLICT OF INTEREST

The authors confirm that there have been no conflicts of interest interfering with the preparation of this manuscript.

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INTRODUCTION

In vivo measurement of loads and displacements in the head and neck is very difficult. For this reason musculoskeletal (MS) models have been useful tools for investigating biomechanical phenomena in this system. For example, MS models offer insight into the relationship between joint loads, muscle lengths and tendon forces during whiplash events which may not be replicated experimentally with human subjects (Brolin et al., 2005; Hedenstierna and Halldin, 2008; Stemper et al., 2004; van Lopik and Acar, 2004; Vasavada et al., 2007). These types of models also have been used to characterize the relationship between computer display heights and gravitational moment due to the weight of the head, muscle moment-generating capacity and other parameters over a range of postures (Straker et al., 2009).

Development of MS models requires several assumptions and simplifications, especially regarding intervertebral kinematics (IVK). IVK may be characterized by the amount of rotation and translation of one vertebra with respect to another, or the amount of rotation and the center of rotation between two vertebrae. In a biomechanical model of the head and neck developed in our lab, the relative motion of each vertebra is assumed to be a pure rotation occurring about a center of rotation fixed in the lower vertebra (Vasavada et al., 1998). Further, the amount of rotation at each intervertebral joint is assumed to be a fixed percentage of the total motion between the skull and T1, and this percentage value does not change over the range of motion. These assumptions make development of head and neck MS models mathematically feasible, but their effects on model estimates are unclear.

Experimental studies have shown considerable variation in IVK among subjects. The distribution of motion among intervertebral segments is found to vary over the range of motion (Anderst et al., 2013b; Wu et al., 2010); for instance, the contributions of the middle cervical levels (C3–C4 and C4–C5) are greater near the neutral posture, but lower cervical levels (C5–C6 and C6–C7) increase their contributions toward the end ranges of motion (Anderst et al., 2013b). In addition, the center of rotation between vertebrae is not constant over the range of motion; it moves anteriorly with flexion movements, especially in the upper cervical spine (Anderst et al., 2013a).

Variation in IVK parameters may affect several MS model estimates. Neck strength, equivalently, the moment generating capacity of the neck muscles, is the sum of moments of all muscles. Muscle moment is the product of muscle force and muscle moment arm, both of which are influenced by IVK. Muscle force is affected through the well-known force-length relationship (Gordon et al., 1966), where muscle length is influenced by IVK. Moment arm can be defined using the tendon excursion method (An et al., 1984), as change in muscle length over joint angle, which is also a function of the amount of motion and location of the center of rotation. Therefore, estimates of neck strength may be influenced by IVK variation.

Moreover, the location of the head center of mass with respect to the trunk or cervical intervertebral joints, is dependent upon the kinematics of each intervertebral joint linking the head to the trunk. Therefore the gravitational load on the neck joints due to the weight of the

head may vary with IVK. The magnitude of this gravitational load relative to neck strength (*i.e.*, the capacity of the neck muscles to oppose the gravitational load), here referred to as *gravitational demand*, is an important MS model estimate for ergonomic applications, (Straker et al., 2009), and this ratio may also be affected by variation in IVK.

The influence of physiological variation on model predictions have been found to be significant in other biomechanical models (Cook et al., 2014), but the influence of physiological variation in IVK parameters has not yet been quantified. The goal of this study was to quantify variation in model estimates of extension strength, flexion strength and gravitational demand (the ratio of gravitational load to muscular capacity) in the sagittal plane due to variation in IVK. Identifying the importance of IVK variation on model estimates is critical for the future application of MS models in evaluating hypotheses related to healthy and pathological functioning of the head and neck.

METHODS

Subjects and Radiographs

Thirty-two subjects with no history of neck pain or prior neck injury were recruited for this study. Approval for this study was obtained from the Institutional Review Board at Washington State University, and all subjects provided informed consent prior to participation in the study. Sagittal radiographs were taken in five postures: maximal extension, mid-extension, neutral, mid-flexion, and maximal flexion as described in a previous study (Zheng et al., 2012). Neutral postures were self-selected by subjects, and maximal postures were voluntarily obtained. Subjects were guided into the mid-extension and mid-flexion postures, which were approximately halfway between the neutral and maximal postures of each subject, as defined by Frankfurt plane angle with respect to the ground. The Frankfurt plane is defined by the line connecting the tragus and the inferior border of the orbital socket.

Coordinate System and Standard Motion Definitions

The location of the right and left tragi and inferior borders of the orbits were marked with lead beads (Y-Spots, Beekley Corporation, Bristol, CT) prior to the collection of radiographs. The corners of each cervical vertebral body, the superior corners of T1 and other anatomical landmarks on the skull (external occipital protuberance, basion and opisthion) were digitized in each radiograph.

The origin of the T1 coordinate system was defined as the mid-point of the T1 superior endplate, with the x and y-axes horizontal and vertical, respectively (Fig. 1A). For each bony structure and in each posture, X and Y and angular position were defined with respect to the T1 coordinate system (relative to horizontal). Skull location was defined as the mid-point between the basion and opisthion, and head angle was defined by the vector from the mean tragus location to the mean inferior orbital socket location (Frankfurt plane), relative to horizontal. The location of the C1 vertebra was defined by the mid-point between the posterior and anterior tubercle, and the C1 angle was defined by the vector connecting those two points, relative to horizontal. Coordinate systems for C2–C7 were defined according to

International Society of Biomechanics (ISB) recommendations (Wu et al., 2002). Locations of these vertebrae were defined by the geometric center (average of the digitized corner points of the vertebrae). C2–C7 angles were defined by the vector originating at the geometric center and orthogonal to the line formed by the mid-points of the superior and inferior endplates (Fig. 1B), relative to horizontal.

For this analysis, we wanted to include only subjects that exhibited standard cervical spine flexion-extension motion. We defined “standard” motion as that generating a “C” shaped curve of the cervical spine, without excessive protraction or retraction motion. We screened subject data for two criteria. First, motion from one posture to the next was considered “standard” if change in head angle with respect to T1 and ground had the same sign. If this were not the case, *e.g.*, if the head were to flex with respect to T1 and extend with respect to ground, it would suggest substantial motion of the entire trunk, motion of T1 within the trunk frame of reference or a combination of the two. Second, motion over a region was considered “standard” if the center of rotation (CR) of the skull with respect to T1 was between the tragus and the origin of the T1 coordinate system. Head CR location was determined using the Rouleaux method (Panjabi, 1979). For example, in “standard” extension, the CR should be located inferior to the skull so that the head moves back and down while extending. On the other hand, for extension with protraction, the CR is located superior to the skull, and the head moves forward and up while extending. Application of these two criteria resulted in further analysis of data for 24 subjects (14F, 10M; Table 1).

Modeling Procedure

The radiographic data collected were used to modify a head and neck model of a 50th percentile male (Vasavada et al., 1998) developed using musculoskeletal modeling software SIMM (Software for Interactive Musculoskeletal Modeling, Musculographics, Inc., Santa Rosa, CA). The IVK obtained from each subject were incorporated into this generic model to create a kinematic-specific model of each subject. The intervertebral kinematic parameters were the X and Y location and angle of each cervical vertebra and the skull relative to the T1 coordinate system. In order to isolate the influence of variation in IVK and reduce the effect of vertebral size and spacing (which would affect muscle length and moment arm), the IVK descriptions of each subject were normalized to the generic model size, as described below.

Because we were interested in motion of the cervical spine only, the orientation of T1 within the musculoskeletal model was fixed with respect to the ground. The generic model had a T1 angle of -23° with respect to horizontal. For the subject-specific kinematic models, the orientation of the T1 vertebra was set as the mean orientation observed in the neutral posture radiographs of the twenty-four subjects, which was -30° relative to horizontal.

For each subject, the X and Y location of each vertebra and the skull were normalized by the T1 to tragus distance (TTD) of the generic male model. We chose the TTD as an overall representation of neck length in the neutral posture. Because TTD can change with head posture, the scaling factor was the ratio $TTD_{\text{model}}/TTD_{\text{subj}}$ at the Frankfurt plane angle (*with respect to T1*) corresponding to the generic model neutral posture. First the TTD_{model} value was obtained from the generic model at a Frankfurt plane angle of zero with respect to

horizontal, which is considered to be neutral head posture of the model. Second, TTD values measured from the five radiographs for each subject were plotted against Frankfurt plane angles (with respect to horizontal), and fit with a cubic polynomial (Fig. 2). The TTD_{subj} value was obtained by interpolating the TTD curve at the skull angle (*with respect to T1*) that corresponded to the generic model at neutral. Because the orientation of T1 with respect to horizontal in the generic model (-23°) was different from the orientation of T1 with respect to horizontal in the subject-specific models (-30°), the interpolation occurred at a skull angle of -7° with respect to horizontal (*i.e.*, the skull angle *with respect to T1* was the same for both subject-specific and generic models). The ratio of $TTD_{\text{model}}/TTD_{\text{subj}}$ was used to scale each subject's respective vertebral and skull X and Y location for each posture.

For each model with subject-specific IVK, motion of each vertebra and the skull over the range of head motion was defined by cubic polynomial fits of anterior-posterior (Tx), superior-inferior (Ty) and rotational (Rz) positions over the head range of motion (Fig. 3). The data from each subject were used to create MS models with unique neutral postures and IVK parameters. All models, however, had the same vertebral size, neck length (in the generic model neutral posture) and muscle parameters. Thus, the only differences between the models were the IVK parameters. This allowed quantification of variation in MS model estimates due only to variation in IVK, without other confounding factors.

Model Estimates

The resulting kinematic-specific models were used to investigate the influence of variation in IVK parameters on estimates of both neck strength (total muscle moment-generating capacity) and gravitational demand. Estimates of moment generating capacity were obtained directly from the musculoskeletal modeling software SIMM. In this modeling environment, the subject-specific kinematics are used to calculate muscle and tendon lengths and the resulting forces in each posture using a Hill-type model of muscle and tendon, where muscles are defined by straight line paths from origin to insertion (Vasavada et al., 1998). Moment generating capacity is the sum of the products of muscle forces and their respective moment arms. We considered both flexion and extension strength over the range of motion. All muscles with extension moment arms were included in the calculation of extension moment, and all muscles with flexion moment arms were included in the calculation of flexion moment. The range of motion was limited to 16 degrees of flexion to 37 degrees of extension so that data from all twenty-four subjects could be included.

Estimates of gravitational demand were obtained by calculating the ratio of gravitational moment (M_g) to neck muscle extension strength estimates (M_m). Gravitational moment was defined as the product of head weight and the distance between the head CR (center of rotation) and CM (center of mass) (Fig. 4, where r_g is the gravitational moment arm, the horizontal distance between the head CR and head CM). The head mass used was 4.5 kg (Yoganandan et al., 2009), and the CM location within the head was estimated according to the definition developed by NASA (NASA, 1978), as 17% of the distance between the tragus and the vertex of the skull. CR location was determined at one degree increments using overlapping postures with head angle four degrees apart.

Gravitational demand $\left(\frac{M_g}{M_m}\right)$ was estimated at intervals of one degrees. The analysis of gravitational demand was limited to flexed postures (0° to 16° of head flexion) because the majority of practical interest in estimates of gravitational demand is related to flexed head postures, which are common in the workplace (Ariëns et al., 2001; Cagnie et al., 2007; Chiu and Lam, 2007).

Additionally, gravitational demand calculations were limited to cases where the head CM was anterior to the head CR, so that gravitational moment was opposed by extensor muscles. As a result, data from two subjects were excluded over a portion of gravitational demand range of analysis, because the head CM was posterior to the head CR in some low flexion angles. Differences in the posture at which the head CM was anterior to the head CR were expected between models as they were created with unique, subject-specific kinematic parameters. Data from one subject were excluded for gravitational demand calculations, because the head CM never became anterior to the head center of rotation location. This was likely due to an extreme flexed torso angle for that subject, so that even when the head was flexed with respect to ground in the radiograph, it was slightly extended with respect to T1.

Muscle Length and Moment Arm

The relationship between IVK and model estimates of strength and gravitational demand is not straightforward; therefore, the observed variation of model estimates to the inter-subject variation in kinematic parameters may be due to a variety of factors. In particular, muscle strength is the product of muscle force and moment arm, both of which are related to muscle length (see Introduction), which is influenced by IVK. Quantifying the relative contribution of moment arm and muscle force to the variation in model estimates is non-trivial due to the highly interdependent nature of the muscle force and moment arm estimates. To remove this interaction and determine the influence of moment arm variation, we also examined model strength estimates when muscle force is fixed to its maximal value. This effectively isolates the effect of moment arm on model estimates, because muscle force is decoupled from the muscle force-length curve.

Statistics

Mean and standard deviations of all model estimates were calculated over their respective range of analysis. Sensitivity to inter-subject variation was quantified using the coefficient of variation (COV), which is the ratio of the standard deviation to the mean.

RESULTS

Neutral posture and changes in head angle over each motion region were characterized by wide ranges among subjects (Tables 2 and 3). Average skull angle in the neutral posture was $5^\circ \pm 7^\circ$, but the range was -9° to 25° . Differences in neutral posture for individual vertebrae (with respect to horizontal) also varied as much as 20 – 30° among subjects (Table 3). Likewise, the total head range of motion from maximum extension to maximum flexion averaged $92^\circ \pm 15^\circ$, ranging from 68° to 118° among subjects (Table 2).

COV of flexion strength estimates (~20 – 80%) were generally larger than COV of extension strength estimates (~8 – 15%) (Figs. 5, 6), though in absolute terms they had comparable standard deviations (approximately ± 5 Nm). Gravitational demand estimates maintained a COV ratio of 80 – 90% (Fig. 7), which was between four and eight times larger than either extension or flexion strength COVs over the flexion range of motion (0 to 16 degree flexion).

The COV for model estimates of extension strength and gravitational demand did not change substantially (less than 2%) when individual muscle forces were held constant at their respective maximal values. This also appeared to be the case for model estimates of flexion strength in flexed postures; but in extended postures, holding muscle forces constant substantially reduced the COV for flexion strength (by almost 50%).

DISCUSSION

In this study, we were able to isolate and quantify the effects of physiological variation in neutral posture and IVK on MS model estimates of neck strength and gravitational demand. These results suggest that more variation is expected in flexion strength than extension strength, and even greater variation (4 – 5 times more) can be expected in the gravitational demand, resulting from the physiological variation in IVK observed in a healthy population. In particular, the gravitational demand is affected by head position, head CR and neck strength, which are all influenced by IVK.

The results of this study were compared to experimental and model strength data over a flexion-extension range of motion. Seng et al. (2002) measured extension and flexion strength in 17 male subjects over a range of 20° flexion to 20° extension. The COV for extension strength varied from 0.22 to 0.28, and flexion strength COV varied from 0.21 to 0.36, with the higher COV values in extended postures. These results are consistent with our findings. The large increase in flexion strength COV in extended postures occurs for two reasons: (1) because of a decrease in flexion strength magnitude (the denominator of the COV ratio), mostly because of decreased flexion moment arms in extended postures; and (2) an increase in the standard deviation of flexion strength estimates in extended postures. When compared to a generic model with kinematic parameters from the literature which do not change over the range of motion (Vasavada et al., 1998), the average of model extension strength prediction with subject-specific IVK are very similar to generic model predictions; the generic model predictions fall within one standard deviation of the subject-specific IVK models (Fig. 5A). Generic model flexion strength predictions are also very similar to the average flexion strength of subject-specific IVK models around the neutral posture. However, beyond approximately 5° flexion or extension, the subject-specific IVK model predictions differ significantly from the generic model flexion strength predictions (Fig. 6A). These results suggest that model predictions of flexion strength, in particular, could benefit from different representations of IVK.

In addition, these results highlight the importance of muscle moment arm estimates on model predictions. In the simulations when force was fixed to its maximal value, the COV in model predictions arose primarily because of variation in muscle moment arms rather than

muscle force-length effects. The COV due to variation in muscle moment arms only was very similar to the COV from both moment arms and muscle force-length effects, except for the prediction of flexion moment in extended postures. Thus, variation in model muscle moment arms due to physiological variation in IVK account for the majority of variation in extension strength and gravitational demand estimates.

The variation in model predictions presented here were based on subjects who displayed a “standard” extension-flexion motion, that is, without excessive protraction or retraction motion. We excluded subjects with head protraction/retraction because we considered protraction/retraction to be a completely different motion from flexion/extension, and we wanted to look only at variation due to flexion/extension motion in this study. Posture of the trunk was not constrained during the experiment, so that the measured kinematics would be more natural, and our measures would be consistent with prior investigations of cervical vertebral motion (Anderst et al., 2013b; Reitman et al., 2004; Wu et al., 2010). Because subjects were asked to assume postures that they could comfortably maintain for the duration of the x-ray, the total ranges of motion observed here were lower compared to previous studies (Demaille-Wlodyka et al., 2007; Feipel et al., 1999; Malmstro et al., 2003; Mannion et al., 2000). As the focus of this work was normal kinematics rather than maximum range of motion, this was not considered a limitation of this work. For subject postures during the x-ray procedure, intermediate postures were defined according to Frankfurt plane angle with respect to ground (because T1 data were not available), but the processed radiographic data are expressed with respect to the T1 coordinate system for more consistency among subjects. Therefore, in a few subjects the change in head angle from neutral or maximum flexion or extension to the intermediate postures (mid-flexion or mid-extension) was minimal (3 - 15°) when angular changes were analyzed with respect to T1 (Table 2).

To assess the influence of trunk posture and Frankfurt plane definition of intermediate postures, subjects with a change in trunk angle more than two standard deviations from the mean for any motion region were identified and models developed from their data compared to the remaining pool of models. Using this approach, four subjects who underwent a particularly large change in T1 angle between postures were identified. Exclusion of these subjects from the pool did not have a substantial effect (less than 10% difference) on mean extension strength, flexion strength or gravitational demand outputs over the range of head motion (Fig. 8A). Likewise, exclusion of these subjects did not substantively affect COV measures (Fig. 8B). While trunk motion undoubtedly influenced IVK, it did not appear to dominate the influence of other factors.

This study was limited in that the radiographs were non-continuous. That is, the subjects were imaged first in neutral, and then in maximal postures before being imaged in the intermediate postures. This was done in order to obtain an estimate of what constituted an intermediate posture for each subject. As a result, the radiographs do not represent measurements of vertebral orientation during a single, fluid movement. Further, intervertebral kinematics used in the models were best fit cubic polynomials of those postures. Intervertebral motion (IVM) measurements at each intervertebral level were compared to data collected from continuous motions for C2–C7 (Reitman et al., 2004; Wu et

al., 2010) as well as data used to define the original generic model kinematics (based on cadaveric data found in White and Panjabi, 1990). The study by Reitman et al. (2004) reported IVM for each segment over the entire head ROM, while the study by Wu et al. (2010) reported IVM for head extension and flexion ROM separately. Our mean IVM measurements were generally within 5% of these literature values (Fig. 9). With the exception of the IVM measures over the extension region of motion, our results displayed the same general trend across intervertebral levels. We observed mean IVM values to be relatively consistent across intervertebral levels over the flexion region, which agreed well with previous observations by Wu et al (2010). Over the extension region, however, we observed more pronounced differences across levels (Fig. 9B), with less relative motion occurring at the C2/C3 and C6/C7 levels than found by Wu and colleagues (2010). In addition to IVM variation, accounting for differences in vertebral translations resulted in altered intervertebral kinematics compared to those assuming a fixed center of rotation.

These results indicate that understanding how different IVK parameters affect muscle moment and head position will be important for predictions of neck strength and mechanical demands on the neck musculoskeletal system. Future work in the development of head and neck MS models should include further investigation of normal and pathological variation in IVK, as well as examining the effects of different modeling assumptions in the definition of IVK.

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Figure 1.

Identified landmarks and the coordinate systems defined in each radiograph. A. Landmarks include the external occipital protuberance (A), left and right inferior orbital socket (B1, B2), left and right tragus (D1, D2), opisthion (E1) and basion (E2), posterior and anterior tubercles of the C1 vertebrae (F1, F2), vertebral corners (G1–G4), and the corners of the T1 superior endplate (H1,H2). The origin of the T1 coordinate system was defined as the midpoint between H1 and H2, with the x-axis horizontal and the y-axis vertical. B. Vertebral coordinate system and orientation definition for the C5 vertebra, as an example.

Figure 2.

T1 to tragus distance (TTD_{subj}) used for scaling subject kinematics. Raw TTD values measured from radiographs for one subject (circles) and the cubic fit of these points (curve) are plotted against skull angle with respect to ground. Estimate of TTD_{subj} (square) was obtained from the cubic fit at the skull angle of -7° with respect to horizontal so that the skull-T1 angle of generic and subject-specific models were the same for estimating this scaling parameter.

Figure 3.

Example of the C6 kinematics from radiographs of one subject plotted against head angle (closed red circles), with the cubic polynomial fit (green curve) and the estimates from the fitted equation (open green circles). A. X position (positive anterior). B. Y location (positive up). C. Z rotation (positive extension).

Figure 4.

Definition of gravitational demand. Moment due to the weight of the head was calculated as the product of head weight ($W_g = \text{Mass}_{\text{Head}} \cdot g$) and the gravitational moment arm (r_g). Gravitational demand was defined as the ratio of the moment due to the weight of the head ($M_g = W_g \cdot r_g$) to the total sum of neck muscle extension moments (M_m ; neck extension strength).

Figure 5.

A. Mean and standard deviation of extension strength estimates of subject-specific kinematic models, and extension strength estimate of a generic kinematic model (Vasavada et al., 1998). B. Coefficient of variation in extension strength estimates.

Figure 6.

A. Mean and standard deviation of flexion strength estimates of subject-specific kinematic models, and flexion strength estimate of a generic kinematic model (Vasavada et al., 1998).

B. Coefficient of variation in flexion strength estimates.

Figure 7.

A. Mean and standard deviation of gravitational demand estimates. B. Coefficient of variation in gravitational demand estimates.

Figure 8.

A. Percent difference in neck strength and gravitational demand estimates when subjects with large trunk motion (more than 2 standard deviations from the mean, $n=4$) were removed. B. Percent difference in COV of neck strength and gravitational demand when those same subjects were removed.

Figure 9.

Mean C2 – C7 segmental IVM values from the present study compared to previous studies and the original generic model (Vasavada et al., 1998). A. Measures over the entire range of head motion by Reitman et al. (2004). B. Measures from Wu et al. (2010) over the extension region of motion. C. Measures from Wu et al. (2010) over the flexion region of motion. Error bars indicate standard deviations.

Table 1

Anthropometric data from 24 subjects included in the study. Data shown are Mean $\pm$ SD (Range).

	All Subjects (n = 24)	Female (n = 14)	Male (n = 10)
Age (years)	24.3 $\pm$ 6.0 [19 – 42]	26.0 $\pm$ 7.3 [19 – 42]	22.0 $\pm$ 2.3 [20 – 26]
Height (cm)	168.3 $\pm$ 11.4 [148.7 – 190.4]	162.1 $\pm$ 6.5 [151.7 – 175.5]	176.9 $\pm$ 11.3 [148.7 – 190.4]
Weight (kg)	23.5 $\pm$ 2.0 [19.8 – 27.4]	22.6 $\pm$ 1.9 [19.8 – 26.7]	24.6 $\pm$ 1.6 [21.6 – 27.4]
Neck Length* (cm)	11.5 $\pm$ 1.3 [8.8 – 14.0]	10.9 $\pm$ 1.1 [8.8 – 12.6]	12.3 $\pm$ 1.2 [9.6 – 14.0]

* Neck length is defined as the vertical distance between the tragus and C7 spinous process when the subject was standing in the upright neutral posture.

Table 2

Change in head angle with respect to T1 between maximally flexed and extended postures (total ROM) and over the extension and flexion motion sub-regions for the 24 subjects included in the analysis. Data shown are Mean $\pm$ SD [Range] in degrees.

Motion Region	Mean $\pm$ SD [Min – Max]
Maximal Extension to Maximal Flexion	92.3° $\pm$ 14.9° [68.4° – 118.3°]
Neutral to Maximal Flexion	37.7° $\pm$ 10.9° [20.6° – 60.0°]
Neutral to Intermediate Flexion	20.4° $\pm$ 9.0° [2.7° – 39.8°]
Intermediate to Maximal Flexion	17.1° $\pm$ 8.8° [5.3° – 44.5°]
Neutral to Maximal Extension	54.6° $\pm$ 11.4° [30.5° – 71.9°]
Intermediate Extension	32.6° $\pm$ 10.2° [15.0° – 52.1°]
Maximal Extension	21.4° $\pm$ 8.1° [7.2° – 35.5°]

Table 3

Neutral posture orientation of skull and cervical vertebrae with respect to horizontal for 24 subjects. Data shown are Mean $\pm$ SD [Range] in degrees.

Segment	Neutral Orientation
Skull	$4.7^{\circ} \pm 6.8^{\circ}$ [$-9.1^{\circ} - 25.1^{\circ}$]
C1	$14.0^{\circ} \pm 5.5^{\circ}$ [$4.3^{\circ} - 25.2^{\circ}$]
C2	$-2.4^{\circ} \pm 8.8^{\circ}$ [$-19.5^{\circ} - 20.1^{\circ}$]
C3	$-8.2^{\circ} \pm 7.6^{\circ}$ [$-22.0^{\circ} - 5.5^{\circ}$]
C4	$-11.8^{\circ} \pm 8.1^{\circ}$ [$-29.1^{\circ} - 2.2^{\circ}$]
C5	$-12.6^{\circ} \pm 7.5^{\circ}$ [$-30.9^{\circ} - 2.3^{\circ}$]
C6	$-13.1^{\circ} \pm 9.2^{\circ}$ [$-36.2^{\circ} - 1.0^{\circ}$]
C7	$-21.7^{\circ} \pm 9.5^{\circ}$ [$-39.9^{\circ} - (-6.1^{\circ})$]
T1	$-30^{\circ} \pm 7.6^{\circ}$ [$-45.1^{\circ} - (-17.6^{\circ})$]