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Passive cervical spine flexion: The effect of age and gender

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ABSTRACT

Background: Previous studies reported passive cervical range of motion under unknown loading conditions or with minimal detail of subject positioning. Additionally, such studies have not quantitatively ensured the absence of active muscle during passive measurements. For the purpose of validating biomechanical models the loading condition, initial position, and muscle activation must be clearly defined. A method is needed to quantify the passive range of motion properties of the cervical spine under controlled loading conditions, particularly in the pediatric population where normative clinical and model validation data is limited.
Methods: Healthy female pediatric (6–12 years; n = 10), male pediatric (6–12 years; n = 9), female adult (21–40 years; n = 10), and male adult (20–36 years; n = 9) volunteers were enrolled. Subjects with restrained torsos and lower extremities were exposed to a maximum 1g inertial load in the posterior–anterior direction, such that the head–neck complex flexed when subjects relaxed their neck musculature. Surface electromyography monitored the level of muscle relaxation. A multi-camera 3-D target tracking system captured passive neck flexion angle of the head relative to the thoracic spine. General estimating equations detected statistical differences across age and gender.
Findings: Passive cervical spine flexion equaled 111.0° (SD 8.0°) for pediatric females, 102.8° (SD 7.8°) for adult females, 103.8° (SD 12.7°) for pediatric males, and 93.7° (SD 9.9°) for adult males. Passive neck flexion significantly decreased with age in both genders ($P < 0.01$). Females exhibited significantly greater flexion than males ($P < 0.01$).
Interpretation: This study contributes normative data for clinical use, biomechanical modeling, and injury prevention tool development.

1. Introduction

Head trauma is the most frequent injury sustained by children (Durbin et al., 2003) and adults (Green et al., 1994; MacKay et al., 1992) in car crashes. Designing effective motor vehicle safety systems to mitigate such injuries requires the use of accurate anthropomorphic models (crash test dummies, human body computer models) to predict the likelihood and severity of head injury in a physical or computer model crash test. The cervical spine of such models governs the motion of the head, as it is the primary structure through which restraint loads are transferred from the torso to the head. This force transfer is influenced by both active and passive neck elements. Though active muscle response is important when predicting the effect of reflex, the inherent electromechanical delay of muscle response suggests that the muscles' contribution to initial head kinematics in unaware occupants may be dependent primarily on its passive mechanical properties. This is supported by a meta-analysis of previous studies arguing that, while neck motion occurred as early as 40 ms after perturbation and injury appeared to occur from 50 to 75 ms, the time required for active muscles to sufficiently influence kinematics was 150–200 ms (Panjabi et al., 1998). Thus, a realistic representation of passive musculature is a necessary feature of a biomechanical model that simulates the early (<150 ms) stages of impact events such as car crashes and contact sports injuries.

Neck model biofidelity is commonly assessed by comparing the model to post-mortem human subjects (PMHS) or human volunteers. Previous studies have quantified passive head and neck motion for adults using whole head–neck PMHS complexes including neck musculature (Cusick et al., 2001; Deng and Goldsmith, 1987; Luan et al., 2000) and passive head–neck computational models have been

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validated using these data (Stemper et al., 2004; Yang et al., 1998). PMHS validation is advantageous because PMHS response can be tested using high severity loading with a direct assessment of injury. The disadvantage of using PMHS muscle as a validation for the passive motion of live human subjects is that postmortem changes (Fitzgerald, 1975, 1977) and the PMHS freezing–thawing process (Gottsauner-Wolf et al., 1995) result in changes to muscle compliance, failure loads, and tensile stiffness. Furthermore, PMHS lack muscle tonus, which even in its passive state influences the maximum ROM (Mertz and Patrick, 1967; Wismans et al., 1987).

In contrast, human volunteer studies provide validation data for anthropomorphic models that include both active and passive muscle effects, but must be conducted at non-injurious levels far less severe than motor vehicle crashes. Previous studies have quantified passive cervical ROM using an examiner to guide the subject's head to full ROM based on the point where a firm resistance was felt (Dvorak et al., 1992; Morphett et al., 2003; Nilsson, 1995; Nilsson et al., 1996; Ordway et al., 1997). Such studies are useful for defining normal maximal limits for detection of cervical abnormalities in the clinical setting. However, in such studies the forces that the examiner applies to the head are unknown making it impossible to validate force-displacement and/or moment–rotation behavior of an anthropomorphic model, and muscle activation via electromyography has not been recorded to ensure muscles are truly passive. Thus, the current study establishes an experimental method for measuring cervical spine flexion under gravity head loading in children and adult volunteers. A novel method to measure and ensure passive neck muscle activation in volunteers is proposed, and the differences in neck flexion in the volunteers across age and gender are presented and discussed.

2. Methods

This study protocol was reviewed and approved by the Institutional Review Boards at The Children's Hospital of Philadelphia, Philadelphia, PA and Rowan University, Glassboro, NJ. Informed consent was obtained from adult volunteers and from a parent/guardian for pediatric volunteers. Informed assent was also obtained from pediatric volunteers.

2.1. Participants

Healthy human volunteers ages 6–12 and 20–40 years with body mass index (BMI) between 10th and 90th percentile were enrolled in the study. Head girth, neck height (opisthocranion to C7), and neck girth were gathered using a flexible tape ruler (Snyder et al., 1977). Head girth was measured at the plane passing from the glabella through the opisthocranion. Neck girth was measured perpendicular to the long axis of the neck at the mid-point. Subjects donned a tight fitting, sleeveless shirt with cutouts to accommodate retro-reflective markers and surface electromyography (EMG) electrodes.

2.2. Instrumentation

Prior to EMG electrode placement, each subject's skin was cleaned (NUPREP, Weaver and Co., Aurora, CO). Disposable, self-adhesive dual surface electrodes (Noraxon, Inc., Scottsdale, AZ) were placed bilaterally on the sternocleidomastoid (SCM), paraspinous (PSP), and trapezius (TR) muscles (Fig. 1A). SCM electrodes were placed mid-belly, PSP electrodes centered along the longitudinal axis of the muscle and placed at the C5 level, and TR electrodes were placed at the mid-point between C7 and the acromion. A grounding electrode was centered over the left mastoidale. EMG data were collected at 1000 Hz using the TeleMyo 2400 T V2 telemetry system (Noraxon, Inc., Scottsdale, AZ).

Cervical spine flexion was quantified using an 8-camera Eagle 1 Digital RealTime motion capture system (Motion Analysis, Inc., Santa Rosa, CA) tracking 10 mm reflective markers at 60 Hz. Markers were placed on the acromion processes bilaterally; external auditory meatus (1.5 cm anterior) bilaterally; head (on tight fitting elastic cap) front, left, right, top; midpoint between xyphoid process and suprasternal notch; nasion; sternoclavicular joint (~3 mm lateral) bilaterally; T1 and T4 (Fig. 1B).

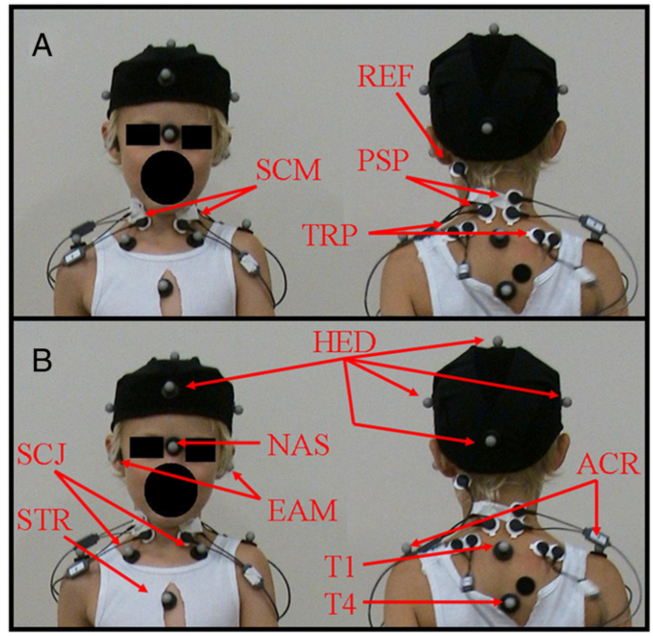

Fig. 1. (A) Electromography (EMG) electrodes were attached to the sternocleidomastoid (SCM), trapezius (TRP), paraspinous (PSP), and a reference (REF). (B) Markers were attached to the acromion processes (ACR), five positions on the head (HED), external auditory meatus (EAM), nasion (NAS), sternoclavicular joints (SCJ), midsternum (STR), T1, T4, and the seatback top and bottom (not shown).

2.3. Testing apparatus

A custom testing device was developed consisting of a rigid seat with padded seating surfaces, four point belt system, and adjustable thigh and lower leg restraints that rotated subjects slowly (10°/s) in the sagittal plane to three pre-defined angles up to 90° (Fig. 2B). An EZ-TILT-2000 rev-2 gravity-based tilt sensor (Advanced Orientation Systems, Inc., Linden, NJ) was placed on the subject's skin between the T1 and T4 markers, allowing researchers to rotate the seat of the test apparatus such that the subject's upper thoracic spine reached specific, predetermined angles with respect to ground.

2.4. Ensuring neck muscle relaxation

To ensure subject neck muscle relaxation, a Passive-to-Active Transition Threshold (PATT) was established for each subject as follows: Seated subjects were instructed to relax their neck musculature, allowing the neck to flex forward under the influence of gravity. Subjects were coached to relax their neck muscles using the phrases "relax your neck muscles, as if you were asleep" and "allow your head to fall forward." Subjects were then instructed to raise their head slowly, and the maximum voltage from any of the measured neck muscles at the moment when the head began to rise was recorded as the initial PATT. This process was iterated and the PATT adjusted until the all voltages were a) just below the PATT in the relaxed state, and b) just above the PATT as the subjects began to raise their head from the relaxed state. The PATT was input into the Noraxon software, providing audio feedback if any of the measured muscles exceeded the PATT during the tests. A smoothing