INJURY BIOMECHANICS RESEARCH
Proceedings of the Thirty-Third International Workshop

Electromyography of Superficial and Deep Neck Muscles during Isometric, Voluntary and Reflex Contractions

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ABSTRACT

The goal of this study was to measure the electromyographic activity of superficial and deep neck muscles during isometric, voluntary and reflexively-evoked contractions. Three male subjects (28 to 41 yrs) had electromyographic (EMG) fine wires inserted into the left sternocleidomastoid, levator scapulae, trapezius, splenius capitis, semispinalis capitis, semispinalis cervicis, and multifidus muscles. Surface electrodes were placed over the left sternohyoid muscle. Subjects then performed i) maximal voluntary contractions (MVCs) in the eight directions (45º intervals) from the neutral posture, ii) 50 N isometric contractions with a slow sweep of the force direction through 720 degrees, iii) voluntary oscillatory head movements in flexion and extension, and iv) initially-relaxed reflex muscle activations to a forward acceleration while seated on a sled. Isometric contractions were performed against an overhead load cell and movement dynamics were measured using 6DOF accelerometry on the head and torso. In all three subjects, the two anterior neck muscles had similar preferred activation directions and acted synergistically in both dynamic tasks. With the exception of splenius capitis, the posterior and posterolateral neck muscles also showed consistent activation directions and acted synergistically during the voluntary motions, but not during the sled perturbation. These findings suggest that the common numerical-modeling assumption that all anterior muscles act synergistically as flexors is reasonable, but that the related assumption that all posterior muscles act synergistically as extensors is not. The data presented here can be used to validate a neck model at three levels of increasing neuromuscular-kinematic complexity: muscles generating force with no movement, muscles generating movement, and muscles responding to induced movement. These increasingly complex data sets will allow researchers to better validate and tune their neck models' muscle geometry, physiology, and feedforward/feedback neuromechanics.

INTRODUCTION

Neck muscles are increasingly included in physical and numerical models used to study traumatic neck injury. Early recognition of the importance of neck muscles on head dynamics resulted in anthropometric test dummy (ATD) response corridors being developed from tests of a pre-tensed human subject (Mertz and Patrick, 1971; Culver et al., 1972). More recently, cadaveric spine preparations used to study traumatic neck injury have also added muscle forces to improve biofidelity (Wilke et al., 1994; Bernhardt et al., 1999; Ivancic et al., 2004). Numerical models have shown that the inclusion of either passive or active neck musculature alters head kinematics during moderate speed collisions (Pontius and Lui, 1976; Williams and Belytschko, 1983; de Jager et al., 1996; vander Vorst et al., 1997). Active neck muscles have also been shown to alter the tensile load distribution in the cervical spine and shift the site of potentially catastrophic injuries from the upper to lower cervical spines (Van Ee et al., 2000; Chancey et al., 2003). Taken together, these studies indicate that neck muscles should be considered when studying traumatic neck injury.

Unlike bones, ligaments and intervertebral discs, musculotendinous tissues are active rather than passive structures. Muscles stiffness varies by an order of magnitude between the passive and maximally-active states (Blackburn et al., 2004; McHugh and Hogan, 2004) and can be transiently increased even further when reflex activation is superimposed on tonically active muscles (Sinkjaer et al., 1988). Muscle activation levels also vary rapidly during a collision: ranging from quiescent levels immediate before and after impact to near maximal voluntary levels within 100 ms of impact (Brault et al., 2000). Moreover, each of the more than 20 pairs of neck muscles (Richmond and Vidal, 1988) – many of which can be subdivided into functionally distinct sub-volumes – may have a different time-varying activation pattern. Thus, the inclusion of neck muscles into a numerical model may improve its biofidelity, but it also increases its complexity.

Current physical models of the neck (e.g., ATDs) include neither individual muscles nor dynamic changes in muscle activation levels. Instead, the neck properties are adjusted to fit within relatively broad corridors (Mertz and Patrick, 1971). Current cadaveric preparations include 2 to 5 simulated neck muscles (Kettler et al., 2002; Ivancic et al., 2004), but do not yet simulate varying muscle activation during the dynamic response. Many numerical models have incorporated more anatomical detail and include 20 or more muscle pairs (Reber and Goldsmith, 1979; Williams and Belytschko, 1983; Luo and Goldsmith, 1991; Vasavada et al., 1998; Van Ee et al., 2000; Chancey et al., 2003; Brolin and Halldin, 2004). Some numerical models include only the passive muscle response (Huston et al., 1978; Yang et al., 1998; Stemper et al., 2004), whereas others use a constant maximal or submaximal activation strategy (Deng and Goldsmith, 1987; Schneider and Bowman, 1978; de Jager et al., 1996; Vasavada et al., 1998; Van Ee et al., 2000) or reflexively-activated muscles (Pontius and Liu, 1976; Williams and Belytschko, 1983; vander Vorst et al., 1997; Brolin and Halldin, 2004). Despite this complexity, most numerical models with active muscles lump anterior and posterior muscles into flexor and extensor groups and then use similar activation intensities for all muscles within a group. Only a few models have varied individual muscle activation levels (Chancey et al., 2003; Brolin and Halldin, 2004).

Validation of these complex neck models is difficult. Typical validation procedures consist of applying an input pulse (often at T1) and then comparing the head kinematics to human subject or cadaver data. Muscle activation levels are rarely based on in-vivo recordings, but instead are tuned either to improve the fit between the model and validation data, or to meet some objective cost function. Neither of these techniques validates the muscle activation pattern used. The goal of this descriptive study is to generate a set of muscle and kinematic data for the validation of numerical models of the neck and its musculature. Three validation levels of increasing neuromuscular and kinematic complexity are considered: isometric (muscles active, but no motion), voluntary head movements (muscle activity generates movement) and impact conditions (externally induced kinematics that are modulated by reflex muscle activation). Kinematic data of the head and torso and electromyographic data of numerous superficial and deep neck muscles are reported.

METHODS

Subjects

Three male subjects with no history of cervical injury, sensorimotor dysfunction or neck/back pain participated in the experiment (Table 1). Subjects gave their informed consent and were paid a nominal amount. The use of human subjects was approved by the university's Clinical Research Ethics Board and conformed to the Declaration of Helsinki.

Imaging

All subjects underwent multiple magnetic resonance (MR) scans to screen for existing pathology, to document neck muscle geometry, and to identify deep vasculature to avoid during wire insertions (Figure 1). Scans were performed with a Phillips Gyroscan Intera 3.0 Tesla magnet using 3 mm slices and a 0.3 mm spacing. Axial T1-weighted scans, proton density (PD) scans and T1-weighted fast field echo (FFE) scans were performed between the T3 spinal level and the occiput of each subject (Figure 1).

Table 1. Subject information.

Subject	1	2	3
Age (yr)	41	29	28
Height (cm)	186	167	180
Weight (kg)	82	75	85
Head circumference (mm)	599		583
C4 circumference (mm)	389	390	389
Neck length (mm)	140	133	131
h (mm) [a]	89	103	85
Frankfort plane angle (°) [b]	19	17	20
Head CoG position (mm) [c]	(42,-136)	(23,-127)	(45,-129)
AO joint position (mm) [c]	(18,-99)	(-1,-90)	(21,-92)

a. estimated vertical distance between the AO joint and line of action of the force applied during the MVC and the isometric sweep contractions (see Figure 3a).

b. while seated on sled.

c. (x, z) position relative to T1 centroid while seated on the sled.

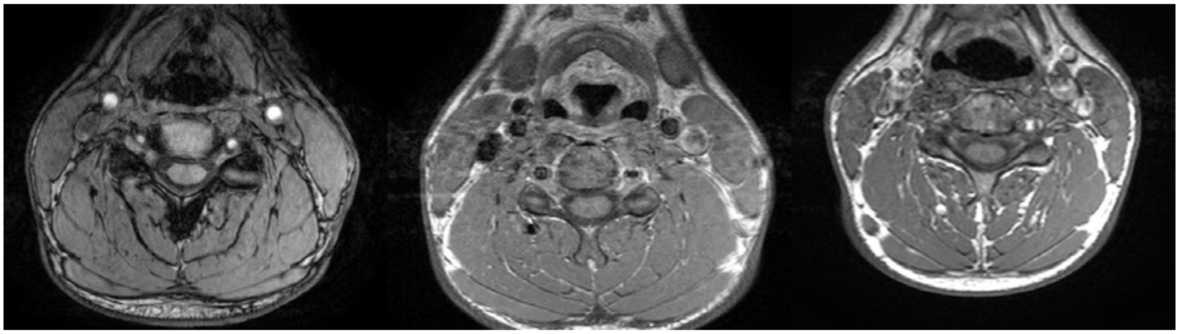

Figure 1: Axial MR images at the C4 level of the three subjects. Shown are: a) a fast field echo (FFE) scan of subject 1, b) a proton density (PD) scan of subject 2, and c) a T1-weighted scan of subject 3. All three types of scans were acquired from each subject and all scans were acquired in the supine position.

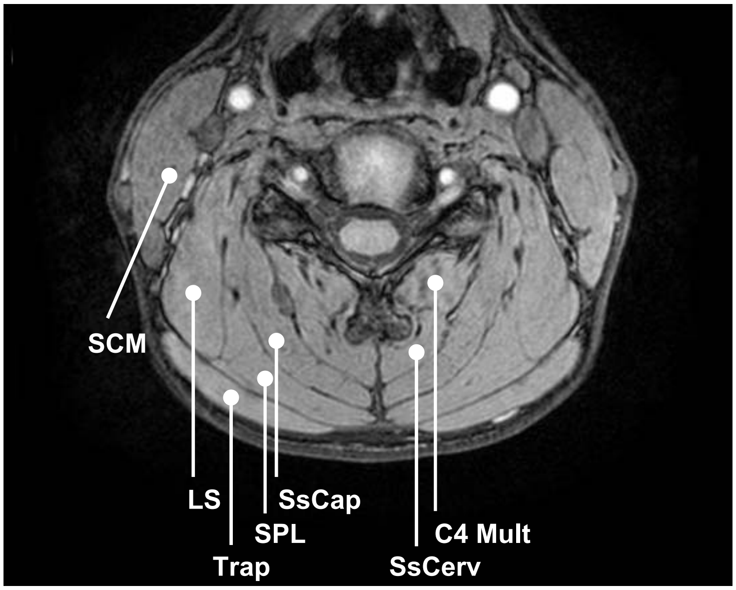

Figure 2: Location of wire insertions into the neck muscles. Insertion shown on right side for clarity only; all wires inserted only on left side.

Instrumentation

Electromyographic (EMG) activity was measured using a combination of indwelling and surface electrodes. Pairs of 0.05 mm wire (Stablohm 800A, California Fine Wire, Grover Beach, CA) were inserted into the left sternocleidomastoid (SCM), trapezius (Trap), levator scapulae (LS), splenius capitis (SPL), semispinalis capitis (SsCap), semispinalis cervicis (SsCerv), and multifidus (Mult) muscles under ultrasound guidance (Sonos 5500, Agilent Technologies, Andover, MA). All wire insertions were placed at the C4/C5 level with an additional multifidus insertion at the C6/C7 level (Figure 2). All wires were placed near the center of the muscle's horizontal cross section. In the SCM, the wire always remained superficial to the readily identifiable cleidomastoid sub-volume (Kamibayashi and Richmond, 1998); During wire extraction, ultrasound was again used to confirm the wire had not moved. Surface electrodes (H69P, Kendall-LTP, Huntington Beach, CA) were placed over the left sternohyoid (STH) muscle with 10 mm inter-electrode spacing. Wire signals were amplified and band-pass filtered (30 – 1000 Hz) using a Neurolog system (Digitimer, Welwyn Garden City, Hertfordshire, England) and surface signals were amplified and band-pass filtered (10-1000 Hz) using a Myosystem 1400 (Noraxon, Scottsdale, AZ) before being acquired at 2 kHz.

Isometric neck loads and moments were measured using a 6-axis load cell (Bertec 4060H, ±10kN vertical, ±5kN horizontal, Worthington, OH; Figure 3a). During dynamic tests, head acceleration was measured with a nine accelerometer array (Kistler 8302B20S1; ±20g, Amherst, NY) arranged in a 3-2-2-2 configuration (Padgaokar et al., 1975). The accelerometers were mounted to a rigid aluminium frame that was securely strapped to a subject's head (Figure 3b). Torso acceleration was measured with a tri-axial linear accelerometer (Summit 34103A; ±7.5g, Akron, OH) and angular rate sensor (DynaCube; ±100 rad/s, ATA-Sensors, Albuquerque, NM). Both torso transducers were fastened to a plate secured immediately below the sternal notch with adhesive and straps. Sled acceleration was measured with a uniaxial accelerometer (Sensotec JTF3629-05; ±20g, Columbus, OH) oriented horizontally along the axis of motion. Displacements were measured with a motion analysis system (Phoenix VZ4000, Burnaby, BC) and nine markers placed on the head, torso and sled. Transducer signals were low-pass filtered (1000 Hz) and then simultaneously sampled at 2 kHz. Displacement data were acquired at 100 Hz per marker. The location and orientation of the head and torso instrumentation was measured relative to anatomical landmarks using a 3D digitizer (FaroArm B08-02, single-point accuracy ±0.30 mm, Lake Mary, FL).

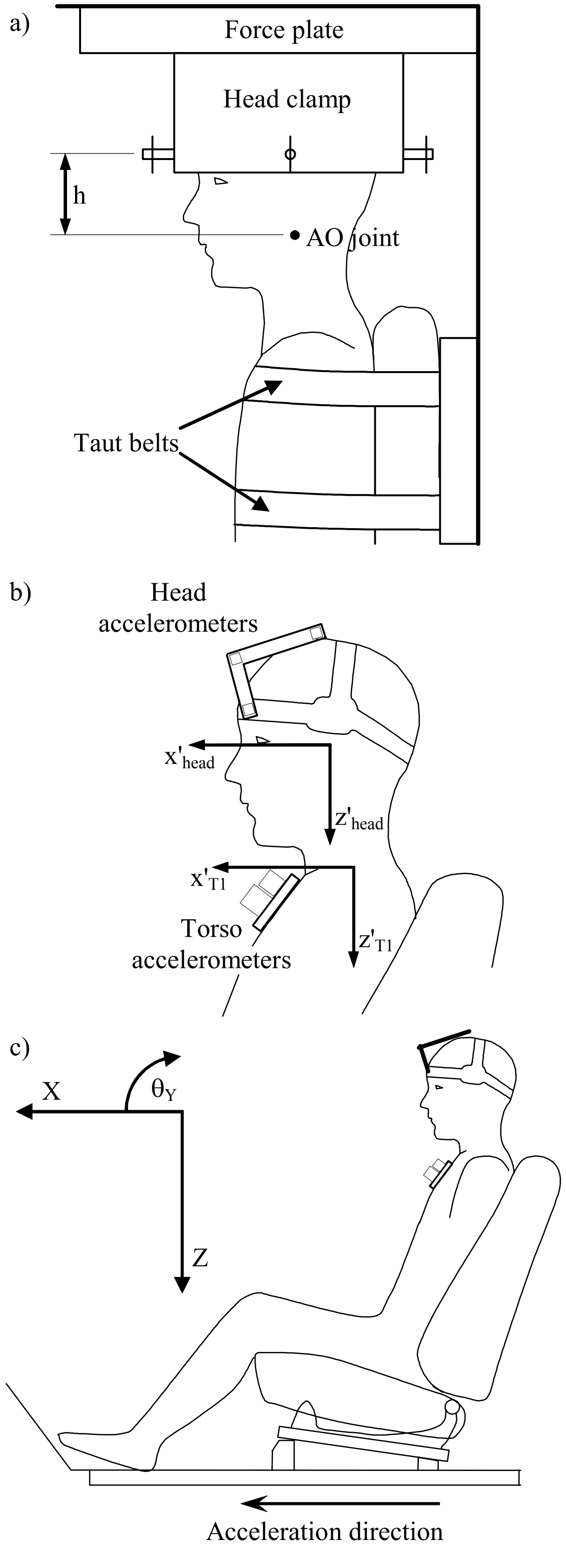

Figure 3. Schematic showing a) the head clamp and force plate used for the isometric contractions, b) the transducers and reference frames used for the voluntary movements and sled tests, and c) the sled configuration.

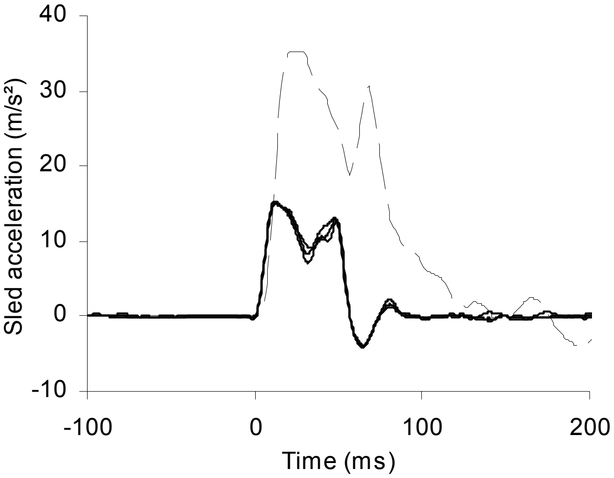

Figure 4: Superposition of the sled acceleration pulses for the three subjects (solid lines) and a vehicle-to-vehicle collision pulse (dashed line) with a speed change of 8 km/h recorded during earlier experiments (Siegmund et al., 2001a).

Procedures

Once instrumented, seated subjects performed four tasks: i) maximal voluntary contractions (MVC) in eight fixed directions (45º intervals), ii) 50 N isometric neck muscle contractions with a sweeping force direction, iii) voluntary oscillatory head movements, and iv) reflex muscle activations to a forward (+x) acceleration. For the MVCs and force sweep tasks, subjects had their head firmly clamped to an overhead load cell and their torso firmly strapped to a rigid seat back (Figure 3a). This arrangement allowed subjects to apply a force in any direction within a horizontal plane nominally at the height of their forehead and opisthocranion. MVCs were held for 5s in each direction and separated by rest periods of 30 to 60 seconds. The eight different MVC directions were presented in an order that minimized fatigue. Subjects received real-time visual feedback of their applied force direction and magnitude (a 2D plot of anteroposterior and mediolateral reaction force) on a computer monitor, and MVC trials in which the force angle was not maintained within 5 degrees of the target angle were repeated at the end of the block of eight trials.

After a rest period, subjects practiced applying a 50 N horizontal force against the load cell and slowly sweeping the direction of the applied force in the horizontal plane through 720º (two complete sweeps, beginning with extension). A 50 N circle was superimposed on the 2D visual-feedback plot of AP and ML forces to assist subjects in maintaining 50 N throughout the entire sweep. Once able to perform the task smoothly, subjects performed four isometric sweep contractions, two each in the clockwise (+z) and counter-clockwise (-z) directions, over a period of about 20s per trial. The middle 360º (about 10s of data) of each trial was subsequently used for analysis.

For the voluntary and reflex-activation tasks, participants were seated in a 1991 Honda Accord car seat with the head restraint removed (Figure 3c). The seat was mounted on a custom-fabricated sled powered by feedback-controlled linear induction motors (Kollmorgen IC55-100A7, Kommack, NY). The sled generated no audible or mechanical pre-perturbation signals that participants could use to predict perturbation onset. Subjects were instructed to adopt a comfortable seated posture, face forward, rest their forearms on their lap, and relax their face and neck muscles. To eliminate habituation effects (Blouin et al. 2003; Siegmund et al. 2003b), subjects received a single forward horizontal perturbation (apeak=1.55±0.03g; tpeak=16.6±0.3ms; Δt=59.0±0.3ms; Δv=0.499±0.003 m/s, Figure 4). After acceleration, the sled continued at constant speed for 0.5s before decelerating to rest. After the perturbation and while still seated on the sled, subjects performed two voluntarily oscillatory flexion/extension neck/head movements. For each movement, subjects started from the neutral rest position and cycled through a comfortable range of flexion and extension movements five times at 1 Hz (a metronome provided a timing signal).

Data Reduction

All wire EMG data were high-pass filtered (50 Hz) to remove motion artifact before calculating the root-mean-squared (RMS) values with a moving 20 ms window. For each MVC trial, a moving average was used to find the 500 ms interval of maximal force and the average RMS EMG for each muscle was then computed for that interval. To generate a normalized tuning curve for each muscle, the RMS value from each of the eight directions was normalized by that muscle's maximum RMS value. For the isometric sweep contractions, the average force and normalized RMS EMG were averaged over a moving 5º interval to generate continuous tuning curves. Moments about the AO joint for the MVCs were computed by multiplying the resultant horizontal force by the moment arm (h) between the AO joint (estimated from the height of the mastoid process) and the line of action of the load cell force (Figure 3a, Table 1).

For the voluntary movements and sled perturbations, rigid body transformations were used to resolve the head and torso transducers into two global reference frames with origins at the head center of mass and center of the T1 vertebral body, respectively. The x-axis was horizontal and positive forward, the z-axis was vertical and positive down, and the y-axis was horizontal and positive to the right (extension was positive about the y-axis). The head center of mass was assumed to be directly above (orthogonal to the Frankfort plane) the midpoint of the inter-aural axis by a distance equal to 17% of the vertical distance between the inter-aural axis and the vertex (NASA, 1978) (Table 1). Based on 41 subjects from a previous study (Siegmund et al., 1997), this CoG position is 23 ± 7 mm forward and 6 ± 5 mm below the position predicted by the regression equations published by Clauser et al. (1969). Based on MR data from the same 41 subjects, the atlanto-occipital (AO) joint in the current subjects was assumed to be 24 mm behind and 37 mm below the head center of mass (Table 1). The center of the T1 vertebral body relative to the manubrium and C7 spinous process was determined from each subject's MR scan. Both external landmarks (manubrium and C7 spinous process) were then digitized with the subject seated on the sled.

All accelerations were determined from the transducers and all displacements were determined from the motion analysis system. All linear accelerometers were corrected for the time-varying orientation of the earth's gravity field prior to computing the six degree-of-freedom kinematics. Peak angular velocity at T1 was determined from the angular rate sensor data after it had been digitally compensated to reduce the sensor's high-pass frequency to 0.002 Hz (ATA Sensors, 2005). T1 angular accelerations were then computed by finite differences (5 ms window) from the compensated angular velocity data. The kinematics of the head relative to T1 were computed by subtracting the time-varying T1 kinematics from the time-varying head kinematics in the lab frame. The angular velocity of the head with respect to T1 was computed by integrating the relative angular acceleration. Retraction was defined as the horizontal rearward translation of the AO joint relative to the moving T1 origin. A subject's initial head and torso position and orientation were determined from the motion analysis markers over a 100 ms period preceding the perturbation and all time-varying kinematic signals were set to zero at the start of the movement or perturbation.

RESULTS

The peak horizontal forces generated during the MVCs varied from a minimum of 118 N in anterolateral flexion to a maximum of 323 N in extension (both in Subject 3, Figure 5). These forces equated to moments about the AO joint of 10 and 29 Nm, respectively. All nine muscles were active over a well-defined directional range between 90 and 180º wide; outside of this range, a muscle was essentially quiet (see bottom four rows, Figure 5). The preferred activation directions for STH (second row, Figure 5) and both multifidus muscles (bottom row, Figure 5) were bilaterally symmetrical, whereas the preferred activation directions for the other six muscles were biased toward the ipsilateral side. Except for SPL, the preferred activation directions of all muscles were similar in all three subjects and consistent with their anatomical position relative to the spine. SPL however, despite its posterior anatomical location, was primarily active during lateral and anterolateral flexion, and on average more than 50% active during pure flexion (thin line, fourth row, Figure 5). In Subject 2, SPL remained more than 50% active during anterolateral flexion toward the contralateral side – a direction opposite to its anatomical position in relation to the spine.

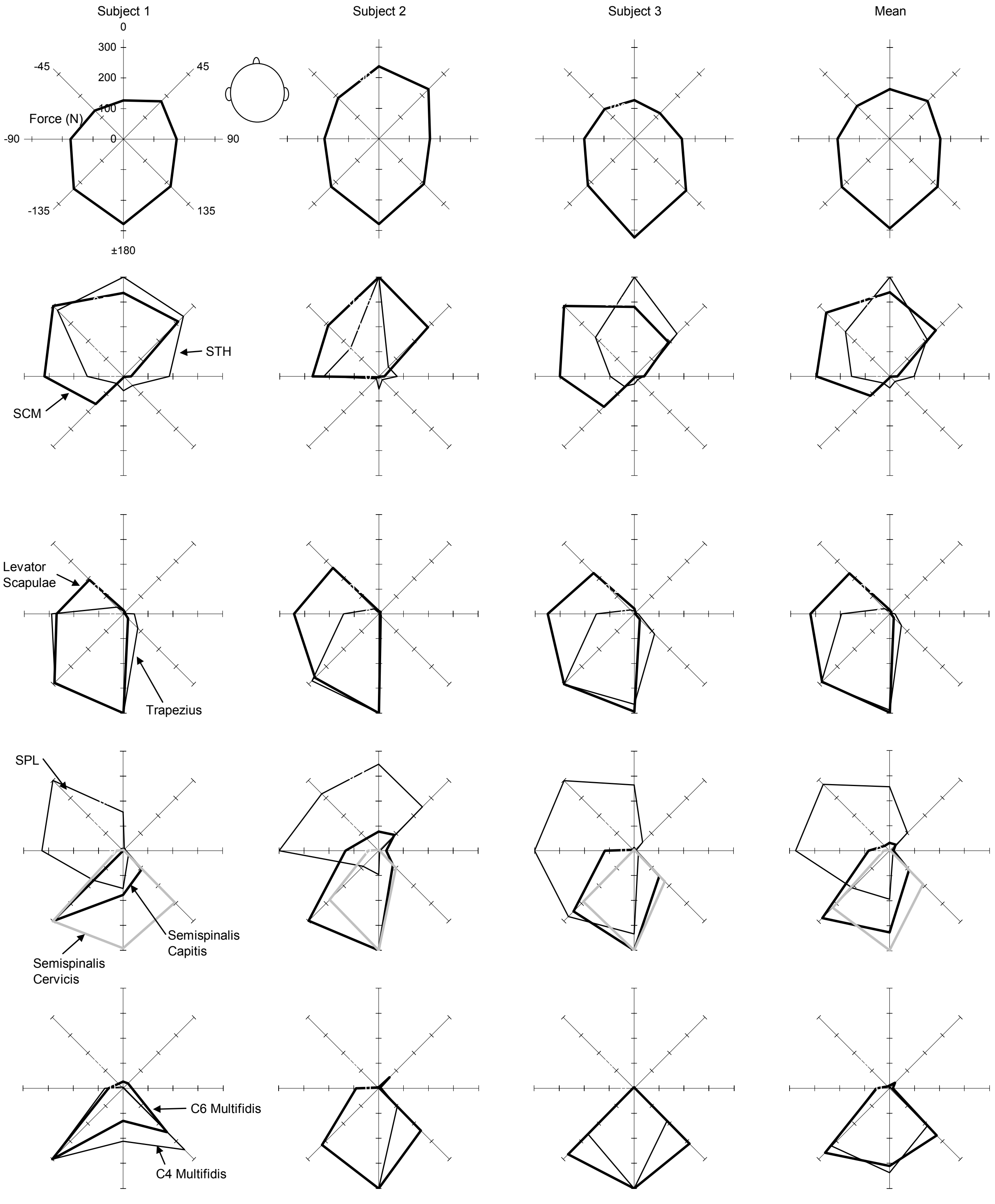

Figure 5: Horizontal force (top row) and directional muscle tuning curves (bottom four rows) for each subject (left three columns), and the mean response of all subjects (right column). Scale and direction shown at top left. All muscle activation levels normalized to the maximum value (100%) for that particular subject's muscle.

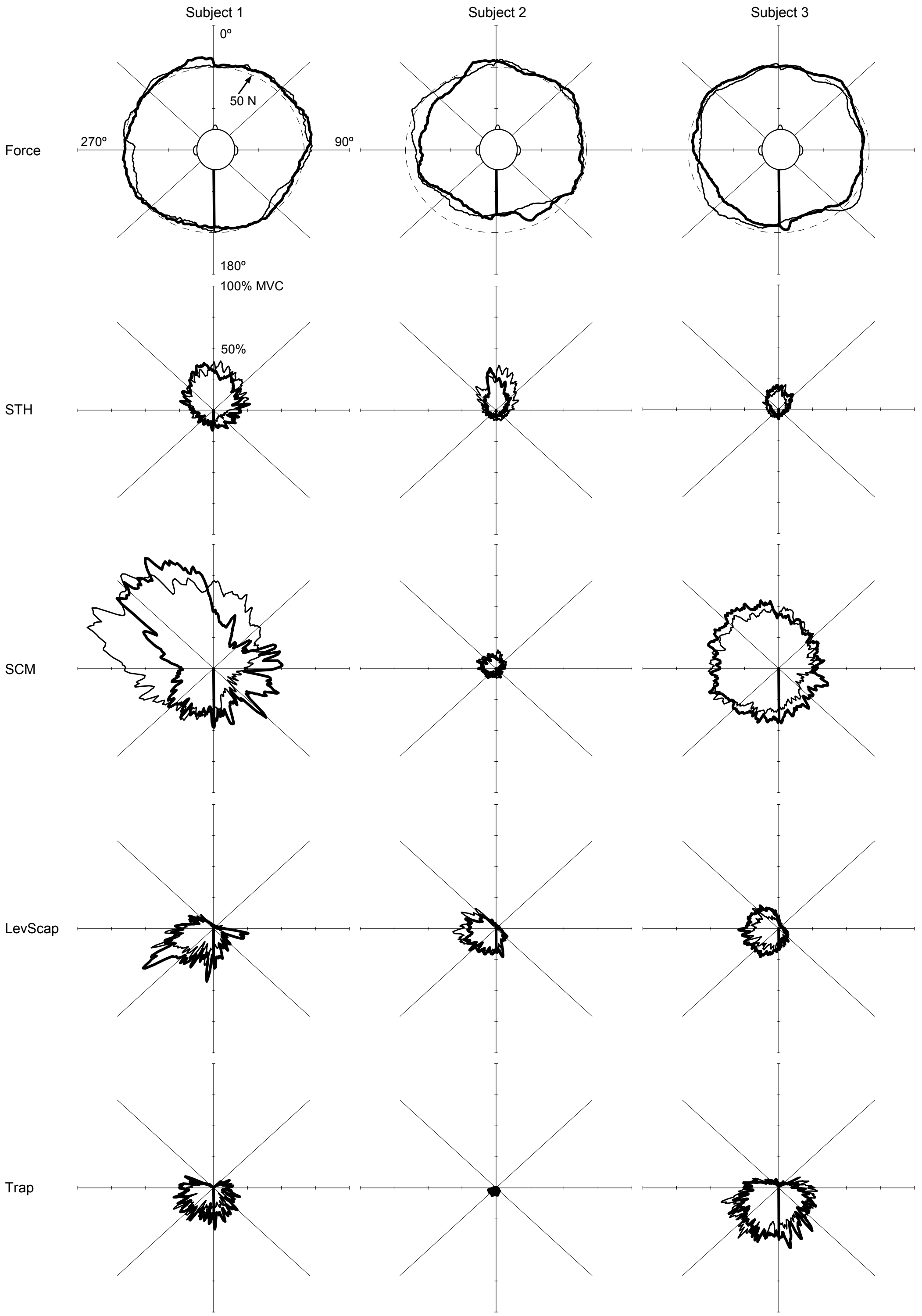

Figure 6: Horizontal force (top row) and normalized RMS EMG recordings (bottom four rows) for sternohyoid (STH), sternocleidomastoid (SCM), levator scapulae (LevScap) and trapezius (Trap) during the 50 N isometric force sweeps. The dashed circle in the force plots indicates 50 N. Clockwise (+z) sweeps shown with a thick line, counter-clockwise (-z) sweeps shown with a thin line.

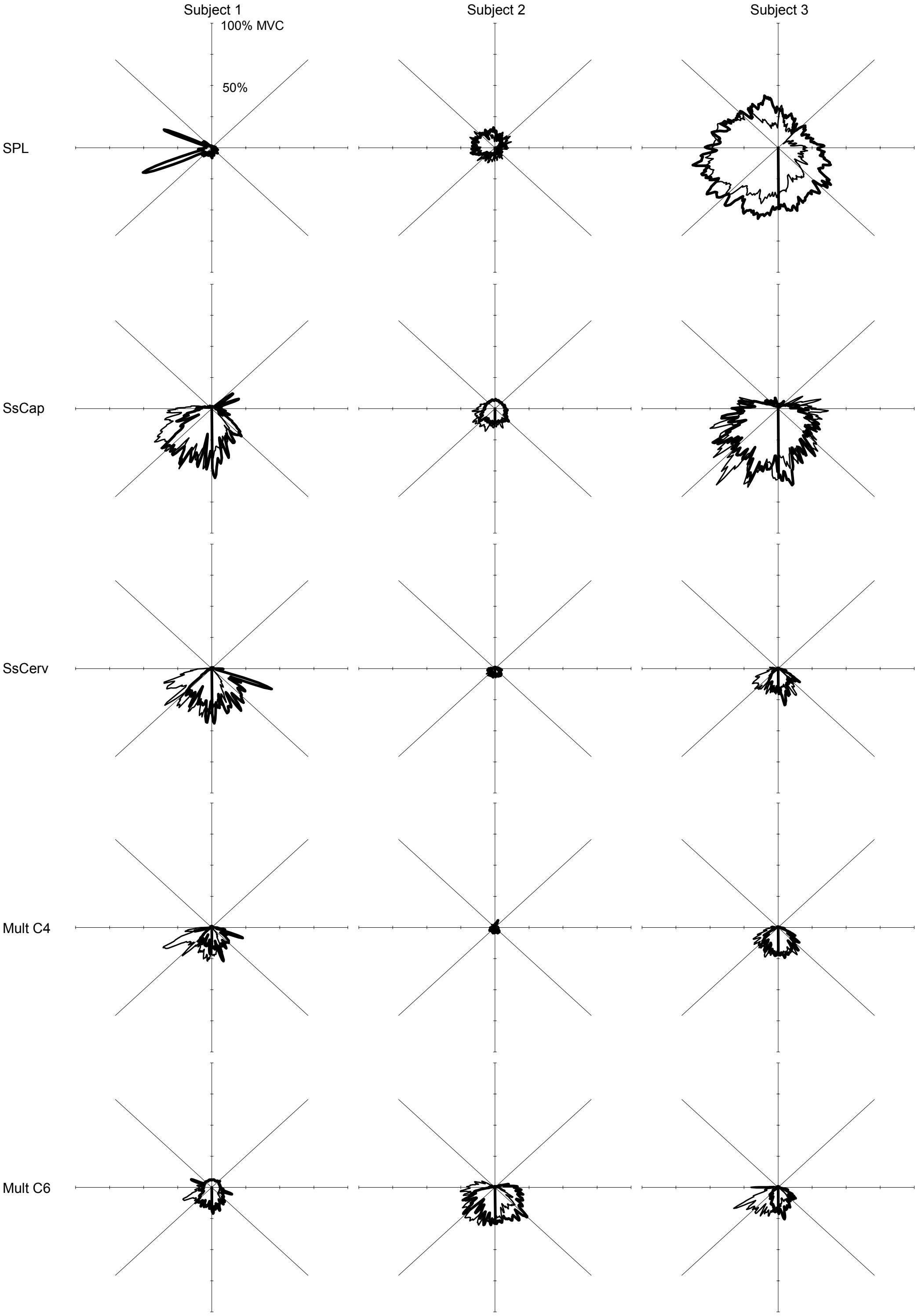

Figure 7: Normalized RMS EMG recordings for splenius (SPL), semispinalis capitis (SsCap), semispinalis cervicis (SsCerv) and multifidus at the C4 (Mult C4) and C6 (Mult C6) levels during the 50 N isometric sweeps (continuation of Figure 6). Clockwise (+z) sweeps shown with a thick line, counter-clockwise (-z) sweeps shown with a thin line.

The 50 N sweep contractions generated muscle tuning curves that were similar (but not identical) to those generated by the MVCs (compare Figure 5 to Figures 6 and 7). Some muscles, particularly SCM in Subjects 1 and 3 (third row, Figure 6) and semispinalis capitis in Subject 2 (second row, Figure 7), remained active throughout the 360º sweep, whereas other muscles, particularly trapezius (bottom row, Figure 6), were active over a larger angular range during the sweep than during the MVCs. All three subjects exhibited similar normalized RMS EMG amplitudes for STH (second row, Figure 6), levator scapulae (fourth row, Figure 6) and C6 multifidus (bottom row, Figure 7). Except for SPL, the normalized RMS EMG amplitudes of the remaining muscles were similar for Subjects 1 and 3, but smaller for Subject 2. Aside from two isolated bursts of activity, SPL remained relatively quiet in Subjects 1 and 2, but relatively active in Subject 3 (first row, Figure 7).

Normalized RMS EMG amplitudes did not vary with sweep direction (compare thick and thin lines in Figures 6 and 7), but some of the clockwise and counter-clockwise tuning curves were slightly rotated with respect to one another. Generally, the angle at which a muscle became active during a sweep in one direction occurred slightly before the angle at which it became inactive in the opposite direction. One exception to this general observation was SCM in Subject 1 (third row, Figure 6), which exhibited elevated activation levels over a wider angle during the counter-clockwise sweep than during the clockwise sweep.

Voluntary flexion/extension movements were generated using a similar strategy in all three subjects (Figure 8). STH and SCM muscles acted synergistically to accelerate the head into extension ($-\alpha$ between the left two dashed lines in Figure 8), whereas semispinalis capitis, semispinalis cervicis and both multifidus muscles acted synergistically to accelerate the head into extension ($+\alpha$ between the right two dashed lines in Figure 8). Levator scapulae and trapezius activity was less consistent – perhaps due to their anatomical association with the shoulder – but levator scapulae did exhibit a large burst during the onset of the first oscillation in Subjects 1 and 2 (third row, Figure 8). SPL again varied between subjects (fifth row, Figure 8): it was silent in Subject 1, exhibited a variable pattern in Subject 2, and was generally in phase with the posterior muscles in Subject 3.

The forward sled perturbation produced similar early head and torso kinematics in all three subjects (Figure 9). Subject motion appeared first as forward (+x) and upward (-z) T1 accelerations and were followed by linear (+x) and angular (+y) accelerations of the head. This lag between T1 and the head produced both neck retraction and combined head/neck extension (θ) (Figure 9). All three subjects had similarly early STH and SCM activation, with peak levels between 1.0 and 2.9 times MVC. This anterior muscle activation preceded peak forward linear acceleration (a_x) of the head with respect to T1 (see dashed vertical lines in Figure 9) and returned to baseline by the time peak head extension occurred about 190 ms after perturbation onset. Posterior muscle activity was also present during the early period of STH and SCM activity, although it was not consistent across subjects. In Subject 1, levator scapulae, semispinalis capitis, and both multifidus muscles were active at the same time as STH and SCM, with both levator scapulae and C6 multifidus approaching MVC activation levels. In contrast, both Subjects 2 and 3 exhibited early activity in SPL, which reached supramaximal levels in Subject 2.

During the rebound motion phase, all three subjects rapidly restored their upright head posture; however, Subject 3 did so without eliminating the retraction that had also developed (see Retraction in the third column of Figure 9). There was generally less muscle activity during the rebound phase than the earlier retraction phase. Semispinalis capitis activity was observed in all three subjects during rebound, with Subject 2 also enlisting his deeper semispinalis cervicis and multifidus muscles at both submaximal and supramaximal levels (middle column, Figure 9).

DISCUSSION

The muscle activation patterns observed during the isometric, voluntary and reflex contractions used in this study showed relatively consistent recruitment patterns for the deep and superficial muscles of the three subjects. In all three subjects, the two anterior muscles – the sternohyoid muscle and the cleidomastoid sub-volume of the sternocleidomastoid muscle – had similar preferred activation directions that were

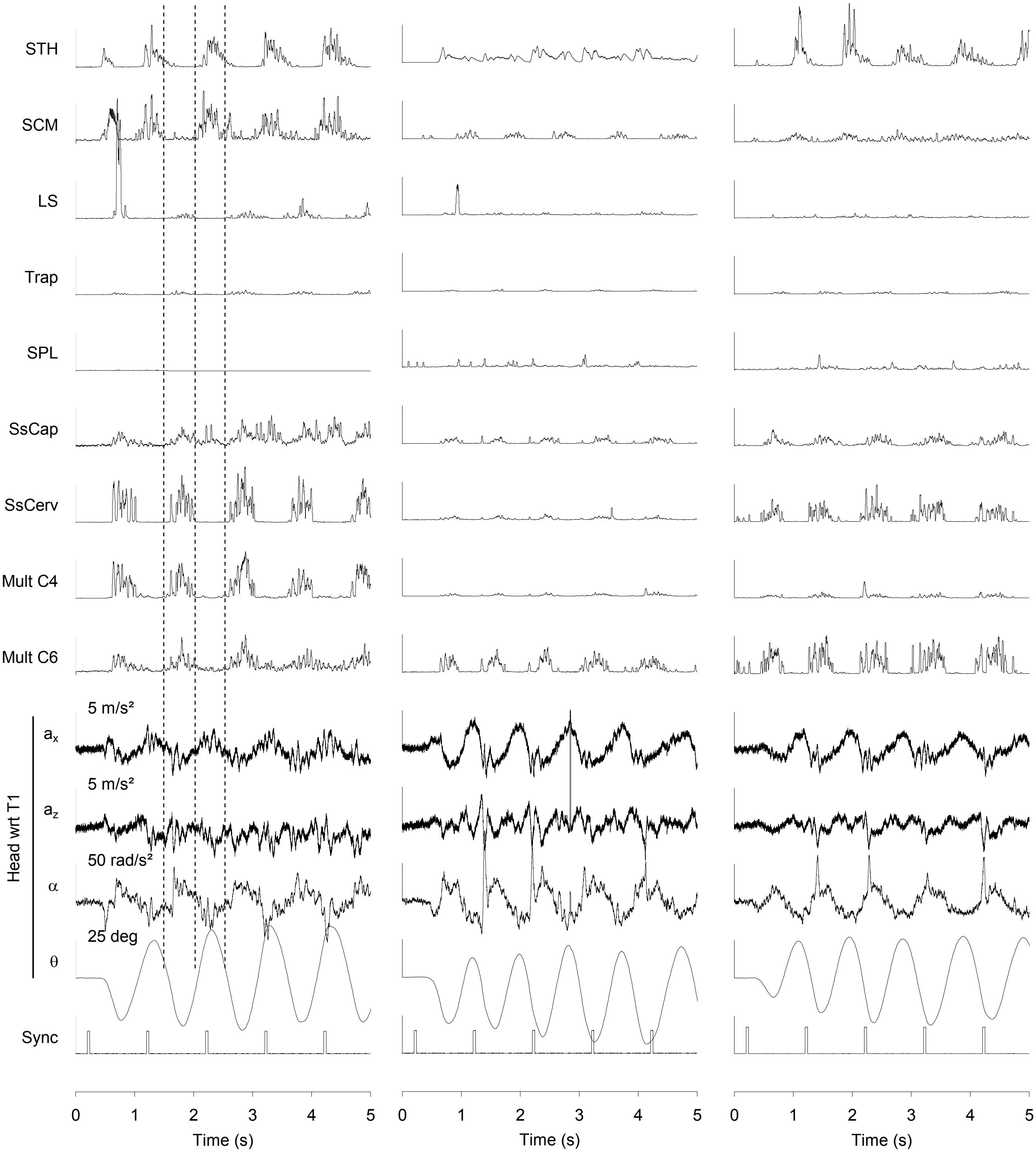

Figure 8: Normalized RMS electromyographic recordings (scale bar = 100%) and kinematics of the head with respect to T1 for voluntary head flexion/extension movements executed by subjects 1 (left column), 2 (middle column), and 3 (right column). Sync refers to an acoustic signal heard by the subjects. (* Increase in baseline noise for SsCap in Subject 1 after sled test but before flexion/extension movement suggests the normalization may be invalid. Use only relative amplitude and timing from this trace). [a, linear acceleration; α, angular acceleration, θ, angle]

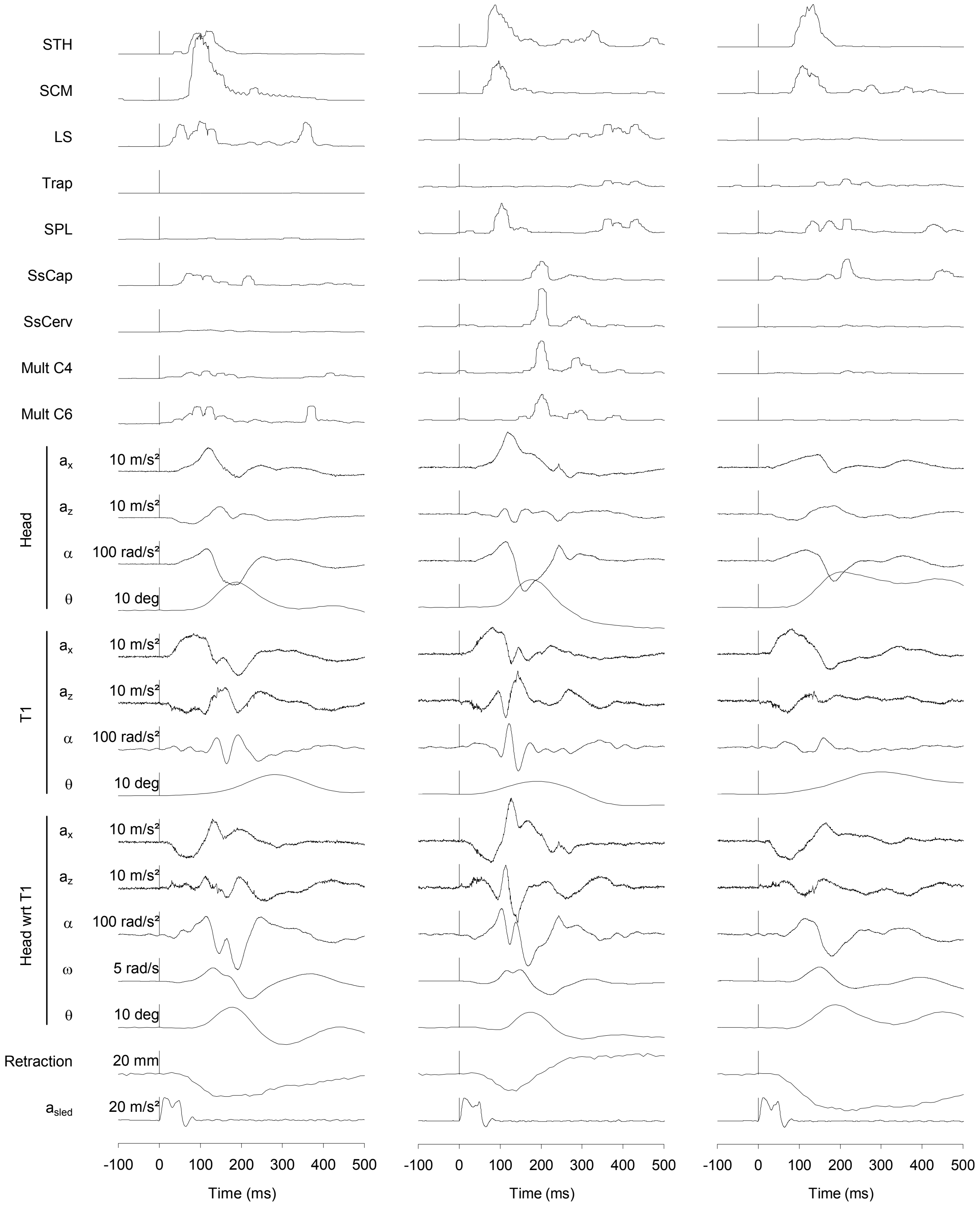

Figure 9: Normalized RMS EMG and kinematics of the head, T1 and head with respect to T1 for forward sled tests experienced by subjects 1 (left), 2 (middle), and 3 (right). Dashed vertical lines aligned with peak horizontal acceleration (a_x) of the head with respect to T1. Muscle scale bars indicate 100% of maximal voluntary contractions (MVCs).

consistent with their anatomical locations. They behaved similarly during both isometric tasks and acted synergistically in both dynamic tasks. With the exception of SPL, the posterior and posterolateral neck muscles also showed activation directions consistent with their anatomical locations and acted synergistically during the voluntary motions. They did not, however, exhibit similar activation strategies across subjects during the sled perturbation.

Of the posterior neck muscles studied, SPL exhibited the most varied behavior. Despite accounting for 30% of the total extensor moment generating potential of the human neck (Vasavada et al., 1998), SPL was most active during lateral and anterolateral flexion, and in one subject active even during anterolateral flexion toward the contralateral side. Previous intramuscular recordings of SPL under isometric conditions have shown either a consistent posterolateral directional preference (Vasavada et al., 2002; Gabriel et al., 2004) or a subject-specific anterolateral and/or posterolateral directional preference (Keshner et al., 1989; Mayoux-Benhamou et al., 1997) for this muscle. Moreover, SPL has a relatively broad insertion along the superior nuchal line, with fibers that insert near the mastoid acting as antagonists with the ipsilateral SCM and synergists with the contralateral SCM to generate or control axial head rotation (Williams, 1995). Thus studies – like ours – that observed both anterolateral and posterolateral direction preferences in SPL may have recorded from more lateral fibers than studies that observed a more consistent posterolateral direction preference in SPL. Since none of the previous studies reported the position of their electrodes within this relatively large muscle, further work is needed, perhaps with multiple wire insertions, to better understand the variable roles (and possibly different compartments) of the splenius muscle.

Despite the inter-subject and inter-study differences related to SPL, this muscle exhibited a consistent directional preference within each one of the subjects studied here. In the subject who used SPL during maximal isometric flexion rather than extension (Subject 2), SPL was active with STH and SCM when a flexor moment was needed to resist extension during the early part of the sled perturbation. In the subject who used SPL during both maximal isometric flexion and extension contractions (Subject 3), SPL was active in both the late retraction and early rebound phases of head motion during the sled perturbation. This within-subject consistency in SPL behavior supports the earlier proposition that SPL may have multiple, functionally-different compartments.

The consistent directional behavior of SPL was not exhibited by all posterior neck muscles studied here. The C4 and C6 multifidus muscles had similar directional tuning curves in all three subjects and were active during angular acceleration in the extension direction (+y) in the voluntary flexion/extension movements. During the sled perturbations, however, the multifidus muscles were active during the initial retraction/extension phase in one subject, during the rebound phase in another subject, and during neither phase in the third subject. The reason for this variable reflex response in multifidus is not clear, but could be related to subject experience: all three subjects had previously experienced the perturbation (> 1 month prior to testing) and knew it was relatively benign. This knowledge may have eliminated an element of surprise (Siegmund et al., 2003a) and reduced the likelihood of evoking a protective co-contraction of the anterior and posterior muscles (Blouin et al., 2003; Siegmund et al., 2003b). More work is needed in inexperienced subjects to better understand this issue.

The neuromuscular and kinematic data presented here have a number of implications for numerical modeling of the human neck. First, the synergy observed between STH and SCM, combined with previous work showing a similar synergy between SCM and the longus colli (Vitti et al., 1973), suggests that the common modeling assumption of uniform activation of the anterior neck muscles appears to be reasonable. Second, the large and consistent activity in STH, combined with a flexor moment arm considerably larger than other neck flexors (e.g., SCM and longus colli), suggests that it should be included in neck muscle models examining sagittal plane motion. And third, the assumption of uniform activation of the posterior neck muscles appears to be incorrect. SPL in particular has a unique activation profile and directional preference, and given its large physiological cross sectional area and a correspondingly large moment arm (Vasavada et al., 1998), likely needs activation dynamics modeled independently of the other posterior muscles.

Numerical neck models are also based on an assumption that muscle recruitment patterns are similar between individuals. While the current data indicate that this assumption is reasonable both for the anterior muscles during the four tasks studied here and the posterior muscles (save SPL) during the isometric and voluntary tasks, it does not appear to hold for the posterior muscles during the sled perturbation and in particular the retraction phase in which whiplash injury may occur. Although multiple strategies for posterior neck muscle recruitment are perhaps not surprising in a system as complex and redundant as the human neck, this finding suggests that modelers should not validate their models by first optimizing muscle activation levels based on some objective cost function and then comparing their optimized muscle activation levels to in-vivo recordings of neck muscle activity. Instead it suggests that neck models need to be validated using in-vivo muscle recordings – possibly from a number of subjects using different recruitment patterns – as inputs, and then comparing the kinematic responses of the model to the human. This latter procedure would allow one to study the effect of specific posterior muscle activation patterns on the overall kinematic response and, in more complex models, the individual tissue responses.

During the dynamic tasks, a number of the muscles studied here exceeded the activation levels measured during the MVCs. Such supramaximal activity has been previously reported in the SCM by Kumar et al. (2002), although the reason for this phenomenon is not clear. One possible explanation is that subjects simply did not exert maximal efforts during their MVCs; however, the neck moments measured in our subjects (minimum 10 Nm in flexion and maximum 29 Nm in extension) are in line with previously reported maximal neck moments (resolved to the mastoid) of 13 ± 3 Nm in flexion and 24 ± 7 Nm in extension (Vasavada et al., 2001). Alternatively, it is possible that some muscles cannot be fully recruited during contractions designed to maximize the horizontal force applied by the head at the level of the forehead. For instance, SCM may require combined flexion and axial rotation to recruit the muscle maximally, although this explanation seems unsatisfactory given that SCM was recruited at supramaximal levels during the sagittal plane motion induced by the sled perturbation. Alternatively, short, reflex muscle activation generates more synchronous bursts of action potentials than produced during a 5 second maximal voluntary effort, and the summation of these synchronous action potentials yields a greater EMG signal. It seems unlikely, however, that this synchronization could explain the 3-fold increase in activation levels observed in the SCM of Subject 1. Whatever the reason, these supramaximal measurements mean that modelers cannot simply use the normalized EMG data to scale muscle force from a muscle's physiological cross sectional area (PCSA). Instead, they may need to first develop individual muscle scaling parameters by modeling the MVC contractions, and then use these scaling parameters during subsequent dynamic modeling. Further work is needed to determine why these supramaximal activation levels occur and how to best contend with them from a modeling perspective.

The currents results also have implications for whiplash injury – particularly injury to the facet capsular ligaments. Recent studies have shown that the cervical multifidus muscles originate from the facet capsular ligaments of the C4-5, C5-6 and C6-7 facet joints (Winkelstein et al., 2001; Anderson et al., 2005). Our data show that in some subjects these deep muscles are active – in some cases supramaximally – during abrupt forward accelerations. Such reflexively-evoked contractions have the potential to increase the strain in the facet capsular ligament above that previously measured under passive loading conditions (Winkelstein et al., 2000; Siegmund et al., 2001b) and thereby increase the potential for injury to this structure. Moreover, the inter-subject variability in multifidus activation levels – from almost no activity in one subject to supramaximal activity in another subject – may also explain why whiplash injury does not develop in all individuals exposed to a particular impact.

In the current study, we limited our analysis to two dynamic conditions that generated head and neck motions in the sagittal plane. The neck can be moved or perturbed in other directions and further work is needed to document the neuromuscular behavior of the superficial and deep muscles under these different conditions as well. We were also limited to eight in-dwelling recordings despite 20 or more muscle pairs in the human neck (Richmond and Vidal, 1988). The muscles we chose represent a large proportion of the muscle tissue at the C4 level, but the contribution of the muscles we did not measure and the contribution of other compartments within the muscles we did measure remain unknown. In addition, only male subjects were presented in the current study, and thus gender differences in neck muscle activation and recruitment pattern remain to be studied.

The speed change and peak acceleration of the collision pulse used in this study is lower than many rear-end collisions. A low severity pulse was used because the head restraint was removed to ensure the observed head kinematics were generated entirely by loads developed in or transmitted through the neck. With both a higher collision severity and a head restraint, peak head kinematics would be driven largely by head-to-head-restraint interaction – a phenomenon which would have reduced the utility of the current data for validating numerical neck models. Despite a lower speed change and peak acceleration, the initial 16 ms of the collision pulse is very similar to larger pulses (speed change = 8 km/h) used in previous studies (Siegmund et al., 1997; 2001a) (see dashed line in Figure 4). Moreover, previous work using surface electromyography has demonstrated a dose-response relationship in muscle activation levels with impacts varying between 1.2 and 8 km/h (Brault et al., 2000; Siegmund et al., 2004), although further work is needed to confirm whether this global pattern applies to each of the deep posterior muscles.

CONCLUSIONS

In summary, the superficial and deep muscle of the human neck generally exhibited consistent directional tuning and activation during the different isometric, voluntary and reflexive tasks used in this study. The anterior muscles were more consistent than the posterior muscles, and the splenius muscle in particular displayed the most variability between subjects and across tasks. Despite these differences, the descriptive data presented in this study can be used to validate numerical models of the neck musculature.

ACKNOWLEDGEMENTS

Thanks to Dr. Marc Carpenter, Mr. Jeff Nickel and Ms. Sophia Hedenstierna for their kind assistance. Thanks also to Dr. Bill Sheel for the use of his ultrasound equipment. Funding for this study was provided by the Canadian Institute for Health Research (CIHR), Michael Smith Health Research Foundation (MSHRF), Workers' Compensation Board (WCB) of British Columbia, BC Neurotrauma Fund, Natural Sciences and Engineering Research Council (NSERC), and University of British Columbia Startup Funding.

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DISCUSSION

PAPER: **Electromyography of Superficial and Deep Neck Muscles During Isometric, Voluntary and Reflex Contractions**

PRESENTER: ***Gunther P. Siegmund, MEA Forensic Engineers & Scientists, School of Human Kinetics – University of British Columbia, International Collaboration on Repair Discovery – University of British Columbia***

QUESTION: *Chris Sherwood, University of Virginia*
Unfortunately, I have had some of this wire EMG testing done and for me it was not comfortable in the least and I'm wondering if you could comment on the discomfort or pain that your subjects may have felt and any effects this may have had on the testing.

ANSWER: Right. The insertion is painful and there's no way around that. You're piercing multiple facial planes in the posterior neck. But once the wires are in and the subject's accommodated to them, we didn't have any reports of pain. We would ask the subjects if they were in pain. They weren't. If someone was experiencing pain and they could isolate it, we'd pull the wire and lose that data channel for that particular test. We did notice a small decrease in the force anterior laterally to the left versus anterior laterally to the right and all our records were on the left. We don't have a large enough end to know if that's significant. You know, it was on the order of 30 or 40 newtons. So the short answer is yes. I mean, you've got foreign bodies in your neck. [laughter] It's going to do something. What? I don't know.

Q: *Guy Nusholtz, Daimler Chrysler*
Have you collected any information on the phase differences between the input, the force inputs or the motion inputs and when the muscles activate?

A: The tracing circles we did counter-clockwise and clockwise so we'd have the information. There is a slight phasing, perhaps less than 5°. It's 10° between the two. So 5° either side. Now those are done very slowly. You know, they'll do 720° over a 20-second period. So I would imagine that as you shorten that time, those phases are going to get longer.

Q: So how much time would you say the response is between an input and muscle response?

A: Are you asking about electrical mechanical delay?

Q: Well, both—One is the physical phase and the other is the electromechanical one.

A: For the isometric and quasi-static stuff, I think we can ignore it. For the voluntary stuff, there's a phase lag. There's some evidence that suggests it's very short, on the order of 15 milliseconds.

Q: 15 milliseconds?

A: 15, yeah. There's some other evidence that suggests that it's as long as a hundred milliseconds. I tend to favor the 15 millisecond, but, you know, it's something we can debate.

Q: Okay. And a comment on your comment about the differences: I'd be interested in seeing if NHTSA could regulate which neck muscles you're going to flex in an accident. [laughter]

A: It's a problem. When you're studying an injury where 5% of the population experiences this and 95 doesn't, you need to test, you know, a hundred people to get five that are relevant. So testing 30, you may not get any that would get whiplash and so you may not be studying the problem.

Q: But, the only reason for the comment is it becomes a very complicated process to go from—Let's assume that that's the info—that that's correct.

A: Yeah.

Q: Then how do you use that information to minimize? I mean, it's an interesting piece of information.

A: Yeah.

Q: Something you can tell people at parties [laughter], but how do you use the information?

A: I don't know. I'm stuck at the beginning of that problem.

Q: *Erik Takhounts, NHTSA*
Yeah, it was quite discouraging. I just want to clarify: You have this multiple activation strategies noted only in dynamic tests, right?

A: There is some variability in the quasi-static and isometric tests, but there's not nearly the kind of variability that you see in the dynamic test and that makes a little bit of sense. We move our head and we probably all adopt similar, near-optimal strategies for regular kinds of head motion. We don't—We aren't often exposed to sled impacts and so we not all have developed the same tuned or optimal responses to those sorts of perturbations. So you would expect more variability in the strategies used for things we experience less than you would for things we experience relatively frequently. I don't know if that's the explanation, but it's one that comes to mind.

Q: I just wonder if the kinematics of the head, for instance, and negative formation patterns like global patterns were similar even though the activation strategies were different.

A: The kinematics would probably generate very nice corridors.

Q: Nice corridors. Alright.

A: Yes, despite the different activation strategies. I guess, it leads to the idea that if you're going to optimize a neck muscle model and you tune your muscles to fit the corridors, you've probably got a tuned muscle system that fits somebody out there but not the general population. So you might need to validate these models the other way around and that is: drive them with the EMG and then look and see.

Q: That basically confirms what we knew before: There is no unique solution for this kind of problem of muscle activations.

A: Yeah. If we don't have a unique solution, certainly the model's not going to have it.

Q: Right. Thanks.

Q: *Mike Kleinberger, Johns Hopkins*
I know you've done some previous work where you've subjected your subjects to multiple tests to look at the learning process. Have you tried to use this process to look at that learning and subject your folks to two or three or more tests to see if you can quantify that?

A: I happen to have a slide!

Q: How convenient!

A: This is a single subject and what we've got here is the same eight muscles during the first exposure and second exposure. So you can see the adaptation that's occurred in some of those muscles. So the habituation that we saw using service electrodes in the past occurs at, as far as we can tell, every one of the muscles that we've taken wire recordings for. So based on a preliminary look at the data, I would say that habituation seems to be a general phenomenon and not a muscle-specific phenomenon.

Q: Great. Thanks.