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Responses of the Trunk to Multidirectional Perturbations During Unsupported Sitting in Normal Adults

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Understanding how the human body responds to unexpected force perturbations during quiet sitting is important to the science of motor behavior and the design of neuroprostheses for sitting posture. In this study, the performance characteristics of the neck and trunk in healthy individuals were assessed by measuring the kinematic responses to sudden, unexpected force perturbations applied to the thorax. Perturbations were applied in eight horizontal directions. It was hypothesized that displacement of the trunk, settling time and steady-state error would increase when the perturbation direction was diagonal (i.e., anterior-lateral or posterior-lateral) due to the increased complexity of asymmetrical muscle responses. Perturbation forces were applied manually. The neck and trunk responded in a synchronized manner in which all joints achieved peak displacement simultaneously then returned directly to equilibrium. Displacement in the direction of perturbation and perpendicular to the direction of perturbation were both significantly greater in response to diagonal perturbations ($p < .001$). The center of mass returned to equilibrium in 3.64 ± 1.42 s after the onset of perturbation. Our results suggest that the trunk sometimes behaves like an underdamped oscillator and is not controlled by simple stiffness when subjected to loads of approximately 200 N. The results of this study are intended to be used to develop a neuroprosthesis for artificial control of trunk stability in individuals with spinal cord injury.

Keywords: sitting stability, trunk stiffness, force perturbation, kinematic response

During unsupported sitting (i.e., without a backrest), the human trunk performs a complex control task in which the muscles of the trunk are contracted in a coordinated manner to maintain an erect posture and resist external perturbations. The intrinsic mechanics of the lumbar spine and its ligaments are insufficient to support vertical loads greater than 88 N (Crisco et al., 1992), therefore the muscles spanning the lumbar vertebrae must contract to support the upper body mass. These contractions are modulated by a combination of feedforward control (tonic contractions, also called stiffness), and feedback control (phasic contractions) in response to external perturbations (Moorhouse & Granata, 2006). The natural control mechanism is considered highly efficient, as it allows for quick, yielding movements in response to perturbations, and a smooth, controlled return to equilibrium. While postural control has been studied extensively, normative values of the response characteristics of the healthy trunk during sitting are not well documented.

The mechanical performance of a control system is typically described by its response to a controlled perturbation. The peak displacement, overshoot, settling time and steady-state error are commonly used in mechanical controller design to assess a control system's stability and performance. The process of optimizing control parameters involves comparing these performance variables to optimal values. In the literature on perturbation studies of postural control, peak displacement is usually the only one of these response parameters reported. Peak Center of Mass (COM) displacement during perturbation is often used as an indicator of postural stability (Bjerkefors et al., 2007; Horak et al., 2005). It is typically presumed that a smaller COM displacement is representative of a more stable system.

External force perturbations have been used in many postural control studies to gain insights into the motor control and stability performance of the trunk (Gardner-Morse & Stokes, 2001; Rietdyk et al., 1999; Stokes et al.,

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2006). Kinematic and neuromuscular responses to perturbation have revealed many characteristics of the control mechanisms of the trunk. Several studies have observed that the trunk response varies with respect to the direction of perturbation (Gardner-Morse & Stokes, 2001; Horak et al., 2005; Preuss & Fung, 2008). This is in part due to the anatomical arrangement of the trunk musculature. The spinal column, which is located in the posterior of the trunk, acts as a fulcrum for flexion and extension. The abdominal flexors have a long moment arm as compared with their antagonists. Whereas the abdominal oblique muscles, which provide lateral stabilizing forces, are arranged symmetrically. We hypothesized that diagonal perturbations, applied in the horizontal plane at a 45° angle to the medial-lateral axis, would result in a less effective response as compared with perturbations in the purely anterior-posterior or lateral directions, due to the increased complexity of an asymmetrical muscular response.

Neuromuscular disorders that affect the trunk musculature, such as Spinal Cord Injury (SCI), often result in postural instability (Bjerkefors et al., 2007; Chen et al., 2003; Horak et al., 2005; Kamper et al., 1999). In a survey of individuals with paraplegia, trunk stability was identified as the third most important gain that would dramatically improve their quality of life (Anderson, 2004). Functional Electrical Stimulation (FES) is being explored as a potential technique to activate the paralyzed trunk musculature during sitting in a way that improves stability and allows individuals with SCI to carry out bimanual tasks, which they otherwise are unable to perform (Kukke & Triolo, 2004; Wilkenfeld et al., 2006). Current FES systems for enhancing sitting stability are experimental and use simple open-loop stimulation of the muscles to stiffen the trunk (akin to anticipatory tonus). These FES systems do not provide reactive responses to perturbations. That is, they are unable to compensate for external perturbations to the torso during FES-assisted sitting. Our primary objective is to develop an FES system that will be able to improve sitting stability in individuals with SCI by providing real-time control over both tonic and phasic muscle contractions in response to an individual's sitting posture and external perturbations. The secondary objective is to develop this system such that it mimics sitting dynamics of healthy individuals. The present study is the first step in that process, which is to analyze and describe the dynamic performance of healthy humans during sitting as a basis upon which to design the proposed FES system.

Our main goal in this study was to collect descriptive data of the normative response to an external force perturbation applied to the thorax. Our analysis focused on characteristics of the time series of COM displacement in response to an impulse perturbation applied to the thorax, and the dissipation of kinetic energy as the body returns to equilibrium.

Methods

Subjects

Thirteen healthy male adults (Age: 21–43 years; Height: 177.0 ± 4.7 cm; Weight: 70.5 ± 9.6 kg) participated in this study. They had no history of neurological disorders or spinal deformity. All participants were right-handed and gave informed consent to participate in the study after receiving a detailed explanation about the purposes, benefits, and risks associated with the participation in the study. The experimental protocol used in this study was approved by the institutional ethics committee.

Apparatus

Each subject was seated on a tall wooden box such that the vertical face of the box was in contact with the calves (see Figure 1). There was no back support and no feet support. The feet hung freely and did not contact the floor. There was light contact between the subject's heels and the vertical surface of the box. This was done on purpose to minimize the contribution of the legs and feet in the control of sitting. No cushions were used. The top surface of the box was a rigid metal plate. The subject wore a tight-fitting t-shirt and a special harness around the thorax. Eight cables were attached to different points on the harness via carabineers. Each cable was used to apply a horizontal force perturbation in one of the following directions, relative to the sagittal axis: 0° (anterior), 45°, 90° (right), 135°, 180° (posterior), 225°, 270° (left), 315°.

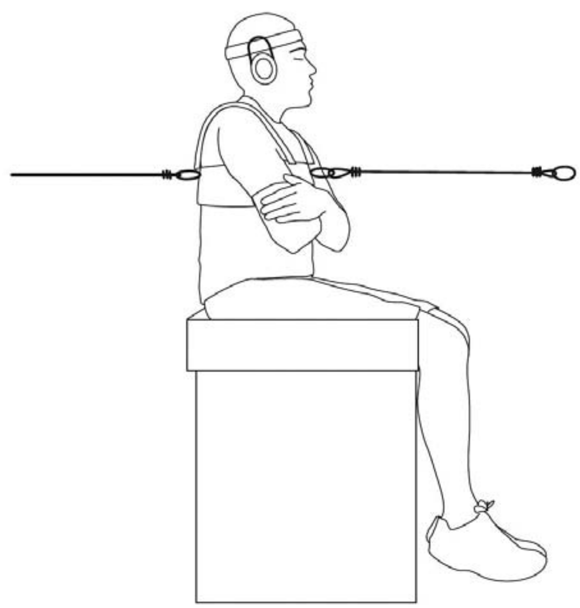

Figure 1 — Illustration of subject sitting on experimental apparatus. Horizontal force perturbations were applied through one of eight cables attached to a harness worn about the chest.

External force perturbations were applied to the cables manually by an experimenter holding one of the cables in series with a force transducer (sensor: MLP-100-CO-C, Transducer Techniques, USA; and amplifier: Model 9243, Burster, Germany). The force transducer had a range of 444.8 N. The experimenter applied a force impulse by quickly tugging on the cable in a practiced manner. The same experimenter applied perturbations in all the trials. All force transducer signals were collected at a sampling frequency of 2,000 Hz using a 12-bit analog-to-digital converter (NI 6071E, National Instrument, USA) and custom data acquisition software.

Kinematic data were recorded using an Optotrak 3020 motion analysis system (Northern Digital Inc., Canada). The 3-dimensional position of 19 markers attached to the subject's trunk and head were recorded at 100 Hz. The marker set used was designed specifically for this experiment. It consisted of seven individual markers located on the left and right anterior superior iliac spines, the left and right posterior superior iliac spines and the spinous processes of C6, T9 and L3. Three clusters of four markers each were mounted on flat square plates. The markers were located at the vertices of squares. The first cluster formed a 50 mm square and was attached to the back of the head with the center of the square located at the opisthocranion. The second cluster formed a 100 mm square and was attached to the back midway between the C6 and T9 markers. The third cluster also formed a 100 mm square and was attached midway between the T9 and L3 markers.

Protocol

The subjects were instructed to cross their arms lightly, close their eyes, and sit in a relaxed and natural posture. A total of 40 perturbation trials (8 directions, 5 trials each) were applied to the subjects at intervals of approximately 30 s. The order of the perturbation directions was randomly determined, such that the subjects were not pulled in the same direction consecutively to prevent anticipation, which has been shown to have a significant effect on the perturbation response (Gilles et al., 1999). The subject wore a headphone and listened to whale music and nature sounds found in national parks. Between perturbation trials, researchers adjusted the ropes to ensure that they were equally slack. This measure was taken to ensure that the subject received no somatosensory cues as to which direction the next perturbation would come. To maintain consistency, all external perturbations were pulled by one researcher. Subjects were offered breaks after every 10 trials. Only one subject requested a break after the 20th trial, which lasted 2 min.

Analysis

A 3-dimensional, 6 degree-of-freedom model of a seated human was developed. The model was an inverted, compound pendulum consisting of three segments: (1) the neck and head, (2) the arms and thorax, and (3) the abdomen. The pelvis and legs were assumed to be static with respect to the inertial frame of reference, and did not move significantly during the experiments. The joints of the model were located at the C6, T9 and L3 vertebrae and were defined respectively as the neck, the thoraco-lumbar joint and the lumbosacral joint. The individual joints had two degrees of freedom, allowing flexion/extension as well as lateral flexion. The 3-dimensional orientation of each segment was determined using three of the markers from each 4-marker cluster. The lower left marker was treated as the origin of the segment coordinate system. The x-axis was defined as the displacement vector between the origin and the lower right marker, the y-axis was defined as the displacement vector between the origin and the top left marker, and the z-axis was defined as the cross product of these two vectors. The top right marker was redundant and was only used when, in some trials, optical markers were not recorded for brief periods of time due to rotation of the marker beyond the angle of illumination or blocking of the line of sight by the cable. In most cases, only one marker of a cluster set was lost, and the orientation of the rigid body could be reconstructed using the other three markers. However, in cases where more than one marker went missing for more than 0.1 s, it was impossible to determine the COM. For this reason, 18 trials out of the 520 recorded were discarded.

The inertial properties of the three body segments were approximated using the regression methods described by Zatsiorsky et al., which are based on total body mass and height (Zatsiorsky et al., 1990). The position of each segmental COM was assumed to be in the geometrical center of the segment, which was estimated using standard anthropometric data (Winter, 1990). Lateral symmetry was assumed.

The following performance variables were determined: (1) The total kinetic energy of the 3-segment model, (2) The peak displacement of the upper body COM in the direction of perturbation, D_x; (3) The peak displacement of the upper body COM in the horizontal direction perpendicular to the direction of perturbation, D_y; (4) D_x normalized with respect to the perturbation impulse—this was done to account for variations in the manually applied perturbation force; (5) D_y normalized with respect to the perturbation impulse; (6) Steady-state error between the initial and final COM displacement; and (7) The settling time, T_s, required for the COM displacement to settle within 5% of the final displacement for at least 2 s. A typical displacement response is shown in Figure 2. The onset of perturbation was determined as the instant when the cable force exceeded 5% of its peak value for each trial. Kinetic energy was normalized with respect to the total mechanical work done by the perturbation, as calculated by load cell measurement and the displacement of the centroid of the four thorax markers in the direction of perturbation.

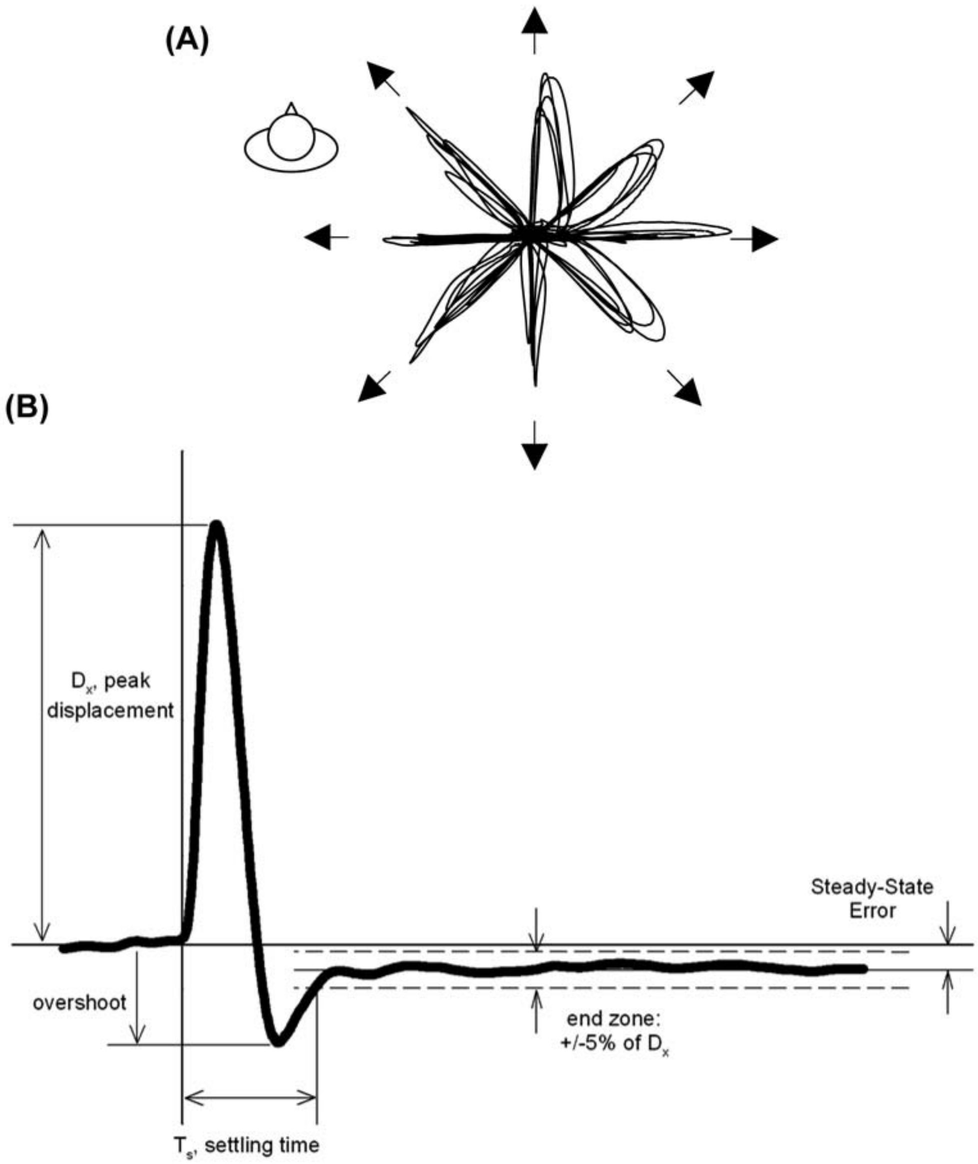

Figure 2 — (A) COM displacement in the transverse plane from all 40 trials (5 in each direction) of subject 1 overlaid. (B) A typical COM displacement response, from which performance variables were identified.

Some trials produced overshoot, which was defined as displacement beyond the final resting position of at least 5% of the peak displacement value. The cases involving overshoot were counted. Vertical displacement of the COM was detected, but neglected in our analyses.

The displacement response was tested with respect to the six performance variables to determine if there was an effect due to perturbation direction or trial order. Hypothesis tests were performed using repeated-measures ANOVA with two within-subjects factors (trials and direction of perturbation). Separate ANOVA models were analyzed for each dependent variable. We interpreted the ANOVA results using the Huynh-Feldt adjustment for violation of the sphericity assumption. Finally, a post hoc analysis was performed to test for linear dependence between mean perturbation force for each subject and the subject parameters: height and weight. Pearson's correlation coefficient (PCC) was determined for force-versus-height, force-versus-weight and force-versus-age. All three PCCs were tested for significance. The level of significance for all tests was set at $p \leq .05$.

Results

The basic kinematic response to perturbations was consistent across all subjects. The neck and thoraco-lumbar joints both flexed/extended in a direction opposite to the external force, while the lumbosacral joint flexed in the direction of the force. Figure 3 shows the individual joint ranges of motion for all subjects, trials and perturbation directions combined.

During the response to perturbation, the three joints of the model flexed in unison reaching their maximum displacements at approximately the same time. This can be seen in the kinetic energy of the response, shown in Figure 4. The kinetic energy reached a peak value of 35 ± 12% of the work done by the perturbation, and then quickly returned to almost zero value, indicating that the body is instantaneously at rest. The kinetic energy then peaked a second time at a much lower value (4.2 ± 0.9% of the work done by the perturbation), indicating a significant dissipation of energy, before returning to equilibrium.

The total upper body COM trajectories for a typical subject are shown in Figure 2B. The COM first tended to displace in the direction of perturbation. Then it sometimes displaced perpendicular to the perturbation resulting in a separate return path. The peak displacement of the COM for all subjects and all trials was 127 ± 42 mm in the direction of perturbation, and 22.6 ± 15.8 mm perpendicular to the direction of perturbation. When normalized with respect to impulse, these values become 3.27 ± 1.11 mm/N·s (parallel) and 0.49 ± 0.32 mm/N·s (perpendicular). Both normalized components of displacement were significantly greater in response to diagonal perturbations compared with perturbations in the orthogonal directions (i.e., anterior, posterior and lateral) ($p < .001$). The settling time was measured to be 3.64 ± 1.42 s, and it did not vary significantly with respect to perturbation direction ($p = .543$). The steady-state error, which is the difference between the initial equilibrium position and the final equilibrium position, was 9.96 ± 8.75 mm, and it did not vary with respect to perturbation direction ($p = .330$). These results are summarized in Figure 5. The displacement response involved overshoot in 105 of the 352 trials analyzed.

The manual perturbation force varied between 92.2 and 293.0 N, with an average value of 186.6 N and a standard deviation of 35.3 N. The perturbation force was found to vary significantly between subjects ($p < .001$) and also with respect to perturbation direction ($p < .001$). There was no significant effect of trial order on the perturbation force, perturbation impulse or any of the performance variables measured. We found no correlation

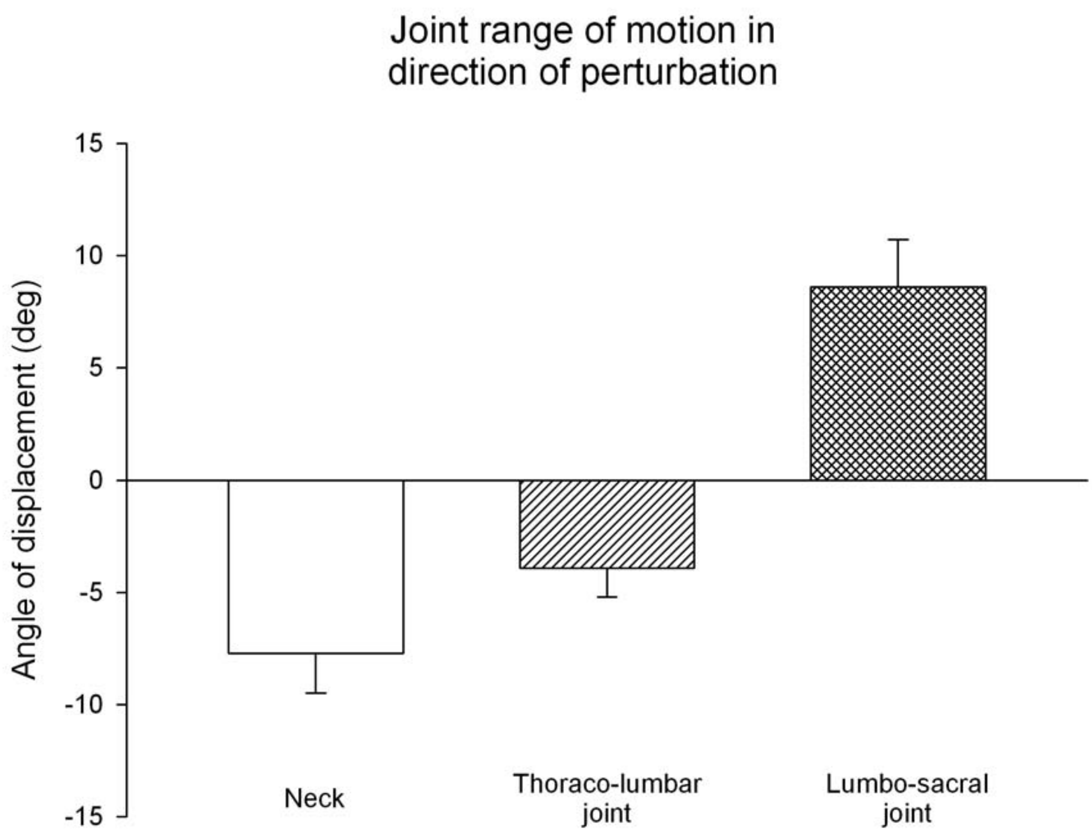

Figure 3 — Range of motion of the segmental joints of all subjects in response to perturbation. Bars indicate mean values. Error bars represent one standard deviation. Negative values indicate joint motion away from the perturbation, e.g., the neck extends in response to forward forces and flexes in response to backward forces.

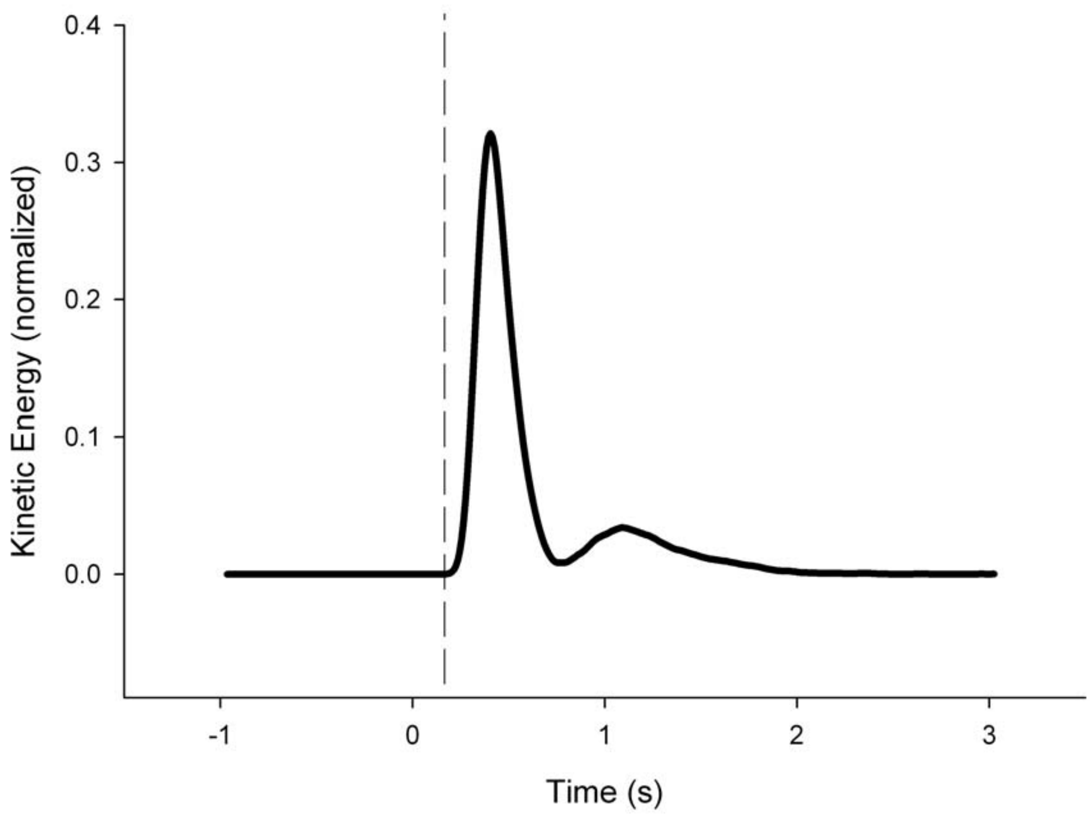

Figure 4 — Kinetic energy response to perturbation (representative sample) normalized with respect to total mechanical work done by perturbation. The dashed line indicates the onset of perturbation.

between mean perturbation force and subject height (PCC = –0.211, $p > .05$) weight (PCC = 0.183, $p > .05$), or age (PCC = 0.073, $p > .05$).

Discussion

Sitting is a very common and important activity of daily living. In certain environments, such as on a train, a bus, or in a wheelchair, unexpected perturbations occur frequently in the form of external forces (bumps and pushes), or movements of the sitting surface. Perturbation studies of postural stability typically use one of these two modes of perturbation: external forces or base-of-support movement. Both are equally relevant to the study of postural stability; however, there are key differences that may not allow for the results to be generalized from one mode to the other. We know of no studies that have investigated the correlation between these two modes. In this study, we set out to evaluate the kinematic responses to short-duration external force perturbations applied to the thorax during sitting. This simulates being bumped unexpectedly, which is a common event in commuter vehicles, classrooms and crowded workspaces. Normative data for the perturbation response to external forces are not well documented. As we wish to design a neuroprosthesis for sitting for people with spinal cord injury, the normal response characteristics to perturbations during sitting are essential for comparing and evaluating the performance of a neuroprosthetic system.

Our analysis of the kinetic energy of the perturbation responses revealed that an average of 35% of the work done by the perturbation was translated into kinetic energy. This indicates an effective damping mechanism. The two-peak profile of kinetic energy during the response indicated that the COM typically came to rest at its peak displacement and then returned to equilibrium at a significantly reduced rate. This behavior is consistent with a damped spring.

The three joints of our head and trunk model acted in a synchronized manner, reaching a state of instantaneous rest. This was confirmed by a near zero value of kinetic energy after the initial response. The body was then returned to equilibrium slowly, the entire response lasting an average of 3.64 s.

Based on the biomechanical and musculoskeletal features of the trunk, we hypothesized that the trunk would perform in a more stable manner in response to perturbations in the anterior-posterior or lateral directions. This hypothesis was confirmed by our observation that perturbation direction had a measurable effect on postural

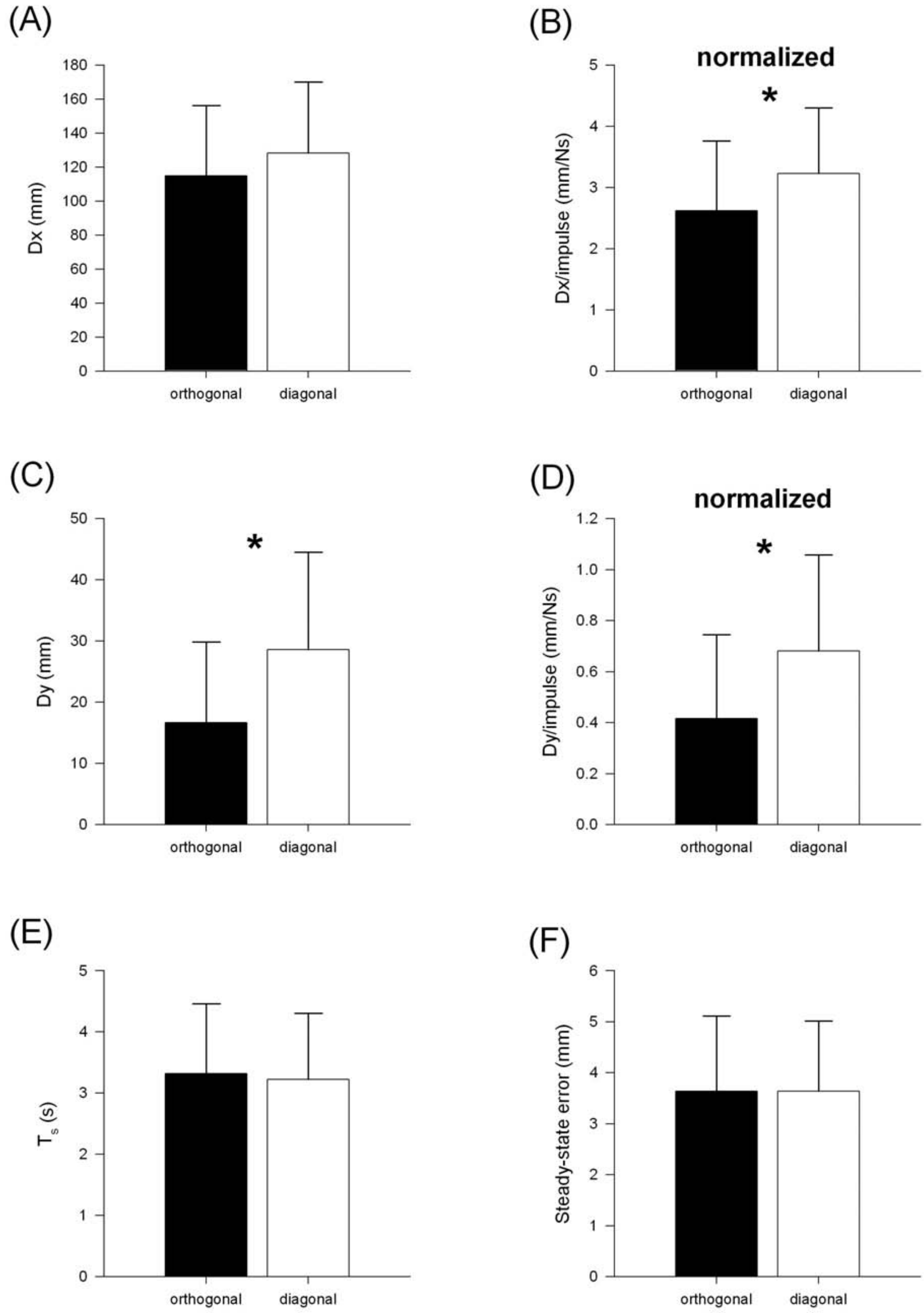

Figure 5 — Comparisons between upper body COM responses to perturbations in the orthogonal direction and perturbations in diagonal directions in terms of the six performance variables. (A) Displacement parallel to perturbation force, (B) normalized displacement parallel to perturbation force, (C) displacement perpendicular to perturbation force, (D) normalized displacement perpendicular to perturbation force, (E) settling time, (F) steady-state error.

stability. This was seen in terms of the peak COM displacement (normalized with respect to perturbation impulse). There was less COM displacement when perturbations were applied orthogonally. This indicates that the trunk is more stable in the purely orthogonal directions. Most of studies that used multidirectional perturbations for standing posture focused on the comparison of body reactions between anterior-posterior and lateral directions (Allum et al., 2003). In standing posture, the body reactions are different, since the dynamics are very different between these two directions, i.e., the ankle joint is dominant in anterior-posterior direction while the hip joint is dominant in lateral direction (Winter, 1990). However, few studies focused on the difference of responses to orthogonal and diagonal directions. Horak et al. (2005) compared subjects with Parkinson's disease to healthy age-matched subjects, and concluded that the healthy subjects demonstrated consistent stability measures in all eight directions, which was contrary to the researchers' expectations.

A significant amount of COM displacement during perturbed quiet sitting was also seen in the horizontal direction perpendicular to the perturbation. This displacement was larger when diagonal perturbations were applied. It likely occurs due to asymmetries in the posture and muscle loads before and during the response. This result was expected, because there is no anatomical symmetry with respect to diagonal axes in the cross-section of the trunk.

Overshoot was seen in 30% of the trials. Overshoot is a characteristic of many physical systems, including an underdamped oscillator or systems with a secondary closed-loop controller. It is not indicative of ineffective control, unless oscillation continues for many cycles. Since it only occurred in a small number of the trials that we analyzed, it is difficult to make any conclusions about normal trunk control from it.

Our analysis of the steady-state error revealed that trunk control is more complex than simple stiffness. Even though the external force returned quickly to zero, the COM rarely returned to its initial position after the perturbation. The difference between the initial COM displacement and the final COM displacement was 9.96 ± 8.75 mm. A system that operates as a simple stiffness controlled system, such as a spring-loaded inverted pendulum, would return to its initial position after the perturbation.

Interestingly, we found no direction effect in terms of the settling time or steady-state error in the COM displacement response. Even though the diagonal perturbations were characterized by larger COM displacements, the COM was returned to a resting position in a similar amount of time and a similar distance from the initial position. This may indicate that the temporal parameters of the reflex mechanisms act to improve upon a simple stiffness response and return the trunk to an equilibrium position by increasing transient muscle forces.

The manually applied perturbation force was not consistent in this experiment. It had a high variability and range. We found that the perturbation force varied significantly between subjects. The force of the pull was not linearly dependent on height, weight or age, so it is not clear why the experimenter pulled harder on certain subjects. We also found that the perturbation force varied significantly with respect to direction (diagonal perturbations tended to be stronger). This is a limitation of the study, and the results should be interpreted accordingly. The increased COM displacement seen in response to diagonal perturbations may be purely an effect of the increased force.

The results of this study reveal some important control mechanisms associated with trunk stability in able-bodied individuals, particularly the effective manner in which kinetic energy is dissipated. Ultimately, these results will be used to develop a neuroprosthesis for artificial control of trunk stability. Our next step will be to implement a simple neuroprosthesis for sitting that uses the open-loop stimulation strategy described in previous studies (Kukke & Triolo, 2004; Wilkenfeld et al., 2006). Subjects with paraplegia due to midthoracic SCI will use this neuroprosthesis. We will measure their response to horizontal force perturbations and compare their performance to the normative values found in this study. Due to the lack of feedback control in such systems, we expect to see less evidence of damping, i.e., increased settling time, frequent overshoot, and greater kinetic energy in the perturbation response. Following this, we intend to develop a closed-loop control system that uses accelerometers to provide feedback for control of the stimulation. The normative values of the response to perturbations reported here will be used to optimize the control parameters of this neuroprosthesis.

Conclusions

This study reveals some relevant characteristics of the control mechanisms of quiet sitting. First, we discovered that the joints of the upper body displace in a simultaneous manner reaching an instantaneous rest position (zero kinetic energy) before returning to equilibrium. However, subjects seldom returned to the initial equilibrium position. Instead, the COM assumed a slightly different equilibrium position. This is an indication that the control mechanism used to maintain sitting posture in response to force perturbation of approximately 200 N does not operate as a simple stiffness controller. We also discovered that the COM displacement in response to perturbation tends to be more pronounced if the perturbation is applied diagonally as compared with perturbations along the sagittal or transverse axes. We observed that direction of perturbation did not influence the settling time or new steady-state position of the COM.

Acknowledgments

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