2004), and measured the experimental ankle stiffness as the linear slope of the ankle torque versus the angle between the vertical and the COG. We then input experimental values of peak ankle torque (Tmax), moment of inertia (I), rocking frequency (ω), and COG height (L) into our model equations to predict the maximum COG amplitude [COGmax = L*(Tmax + Iω^2 * θmean)/(Iω^2 + mgL)] and ankle stiffness [k = ω^2I + mgL] at each frequency.

Results: The measured COGmax averaged 11.9 cm (SD 1.6) at 0.33 Hz, and decreased to 9.2 cm (SD 1.2) at 0.66 Hz. Model predictions of this parameter averaged 11.9 cm (SD 1.3) at 0.33 Hz (accurate to within 0.1%, on average), and 9.0 cm (SD 1.2) at 0.66 Hz condition (accurate to within 2.9%). The measured subject-specific ankle stiffness based on the full range of torque-rotation data averaged 687 Nm/rad (SD 118) at 0.33 Hz, and 1304 Nm/rad (SD 203) at 0.66 Hz. When data were omitted corresponding to the outer quartiles of displacement (where stiffness varied nonlinearly), this slope increased to 787 Nm/rad (SD 156) at 0.33 Hz, and 1689 Nm/rad (SD 243) at 0.66 Hz. Model predictions provided a strong match to the latter, averaging 788 Nm/rad (SD 152) at 0.33 Hz (accurate to within 0.01%), and 1637 Nm/rad (SD 296) at 0.66 Hz (accurate to within 3.4%).

Conclusions: Heel-toe rocking represents an idealized COG-displacing activity that nevertheless may mimic the essential dynamics governing balance maintenance during activities such as reaching. The dynamics of this task are simulated remarkably well by the inverted pendulum model, as evidenced by (1) the accuracy of the stiffness equation in describing the measured torque-rotation behaviour of the ankle, and (2) the ability of the model to predict the effect of movement speed on postural stability borders. These results encourage the development of more comprehensive assessments and corresponding mathematical models based on this task.

SP-30

Modeling of multi-joint postural control based on kinematic and EMG data

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Introduction: From a control theory perspective, the postural control system consists of two processes: the mapping from muscle motor commands to sway (the plant) and the mapping from sway to muscle motor commands (feedback), where we consider rectified EMG activity as an proxy for muscle motor commands. Using a linear approximation, each of these mappings can be characterized by an open-loop frequency-response function (FRF). Fitzpatrick et al. (1996) used sensory (galvanic) and mechanical perturbations to identify the plant and feedback FRFs based on a single-joint (ankle) model of the body. We extended this method by using visual perturbations to identify the plant FRF assuming a double-joint (ankle and hip) model of the body. We then used the identified plant FRF to develop a mechanistic multi-joint model of the plant.

Methods: Subjects stood surrounded by front, left and right screens simulating a visual scene rotating about the axis through the ankles. The rotation signal consisted of ten sinusoids ranging in frequency from 0.024 to 2.936 Hz. Trunk and leg angles in the sagittal plane and rectified EMG signals from soleus, gastrocnemius, tibialis anterior, rectus femoris, biceps femoris, erector spinae and rectus abdominus muscles were analyzed. We computed closed-loop FRFs from visual scene position to each segment angle and EMG signal.

Results: The gain from visual scene position to EMG activity were qualitatively similar for all muscles. Cophases were similar across frequency for all posterior muscles and for all anterior muscles. Cophases for posterior and anterior muscles differed by approximately 180 degrees. Given this roughly fixed relationship between muscle activity across frequency, we modeled the plant as having a single input. The input was the weighted sum of all EMG signals, with posterior and anterior muscles assigned positive and negative weights, respectively. Cophases from visual scene position to the leg and trunk angles were similar at low frequencies and approached a difference of 180 degrees at higher frequencies. Therefore, we considered leg and trunk angles as separate plant outputs. We identified the SIMO (single-input multiple-output) plant by dividing the the FRFs from visual scene position to segment angles by the FRF from visual scene position to combined EMG activity. We modeled the plant using a two-joint (ankle and hip) model of the body and a second-order low-pass transfer function for the mapping from EMG activity to joint torques.

Conclusions: The plant model was in general agreement with the empirically identified plant FRF. We are currently combining the plant model with various models of feedback. In these posture models, the change in leg-trunk coordination across frequency is due to properties of the plant and are not produced by the feedback control strategy. Supported by NIH grants RO1NS35070 and RO1NS046065.

SP-31

DYNAMIC SIMULATION OF THE BEHAVIOR OF AN ORTHOSIS FOR KNEE AND ANKLE FUNCTIONAL COMPENSATION DURING GAIT

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Introduction: An intelligent orthosis for knee and ankle joints functional compensation has been developed in the frame of the European project Gait (EU contract IST-2001-37751). Such a device can be worn by patients with an abnormal gait caused by muscle weakness due to neurological and muscular diseases in order to provide safety and walking pattern improvement. The idea under the design of this system was to approach the behavior of each joint, knee and ankle, as a linear elastic spring, constructing an actuating system to be the substitute of the real muscles. According to this approach, different springs will provide the needed actions for each joint. Selection of those springs is done based on two parameters, patient's weight and walking speed, and according to the hypothesis that joint angles will be similar during gait for a patient wearing the orthosis and for normal subjects. To evaluate this hypothesis it was necessary to test the concept previously with a computer model. This paper presents the computer model with all assumptions and simplifications used in order to simulate the behavior of the orthosis-leg system during the gait cycle.

Methods: Software used for the simulation was ProEngineer Wildfire 2.0. The orthosis was modeled as a mechanism formed by different bodies with the same kinematic relationships as the real orthosis. The leg was modeled as a passive mass consisting of three bodies, thigh, calf and foot, being each one rigidly attached to the corresponding part of the orthosis. Steel, aluminium, and carbon fiber densities were considered for the orthosis, and a density of 950 Kg/m3 was considered for the patient leg. All bodies were supposed to be rigid. In order to model the actuators, elastic constants of the active springs in each case were introduced. In addition, in order to simulate the performance of the system during gait cycle, stance and swing were studied separately. Stance movement was divided into three phases, considering two or three degrees of freedom depending on simulation conditions. Hip angle, knee angle and foot angle with the floor were used as DOF's, being the last one restricted while the foot leans completely on the floor. Data from hip horizontal and vertical reactions and torque applied were used as inputs. Swing movement was split into two phases, the system was supposed to have three DOF's (Hip, knee, and foot angles), and the hip position was fixed. Torque applied on the hip was used as input.

Results: Despite some differences with the theoretical gait pattern, kinematic and dynamic results showed that the hypotheses and simplifications used during the design of the actuating system are valid. Moreover, the model was revealed as a useful tool in order to adapt the constants of the springs in the orthosis to each patient since different values can be evaluated.

Conclusions: The developed computer model can also be used as starting point for future computer simulations of gait cycle, with or without orthotic devices.

Cognitive Influences on Posture and Locomotion

SP-32

Exploring the process interference between balance control and working memory by an event-based analysis of body sway microstructure

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Introduction: Previous studies have demonstrated interactions between balance control and cognitive processes. However, simultaneous cognitive load does not always impair stability of balance. Possible causes for this inconsistency of effect include uncontrolled performance trade-offs between the two task domains and possible confounding between difficulty and process complexity of the cognitive task.

Methods: The present study develops a new methodological approach to avoid these issues by analysing the microstructure of body sway related to immediately preceding stimuli in a concurrent cognitive task. While standing with eyes open in narrow base stance, 10 adult volunteers were tested in single and dual-task mode. In the dual-task condition, subjects performed a sequential 1-item numerical memory updating task presented on a computer screen. The difficulty of this cognitive task was parametrically varied by adaptively adjusting the exposure time to define a performance operating characteristic for each subject. In order to keep performance trade-off constant across difficulty levels, the difficulty level for each following trial was selected at random. Participants' ground reaction forces were registered using Bertec forceplate to yield centre of pressure (CoP). Postural "threat" at each point in time was calculated from CoP using the estimated time-to-contact (TtC) the stability boundary. State-dependent intervention by the balance control system was represented as increasing TtC rate (dTtC). Assuming that the effect of any intervention of the balance control system varies in time, the direction specific dTtC distributions were determined. For segmented regions within the stability boundaries, differences in the dTtC distributions were calculated as a function of the dual-task condition and the level of postural threat.

Results: Preliminary analysis of the data from a single subject revealed that, in the single task condition, close to the stability boundary the dTtC distribution was skewed towards the central point (directionally specific balance adjustments). In contrast, in the central region of the stability range, the dTtC distribution was directionally non-specific, indicating reduced levels of postural intervention. There was no difference between the single task and dual-task conditions in the dTtC distributions for the central region but a difference was apparent in the region close to the stability boundary. Its dTtC distribution lost directional specificity under dual-task condition indicating the involvement of working memory resources in state dependent postural adjustments.

Conclusions: Event-based analysis of sway microstructure thus provides a means of demonstrating that concurrent cognitive demands on working memory affect the distributions of direction specific TtC adjustments. We interpret our results in terms of competition between balance and cognition for processing resources involved in the planning and execution of balance adjustments.

SP-33

Spinal postures in sitting: can researchers and subjects differentiate postures at lumbar and thoraco-lumbar regions?

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Introduction: Since the 1950's an 'ideal' spinal posture in standing was proposed to involve a slight lordosis at the lumbar and slight kyphosis at the thoracic spine. These spinal curves have also been advocated clinically as an 'ideal' sitting posture. Surface measures of spinal posture lack a clear standard for methodology, and few studies have quantified spinal curves in upright postures. Our first objective was to quantify sagittal spinal curves in sitting, using surface measures at thoracic, thoraco-lumbar