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MECHANISMS AND MODELS OF POSTURAL STABILITY AND CONTROL

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Abstract— **Though simple in appearance, postural stabilization is a complex neuromuscular task requiring coordination among multiple joints. Mechanisms of postural stability and control in the body include supraspinal processes responsible for anticipatory postural adjustments (APA) and internal model control, lower level motor servo, and passive viscoelasticity of the musculo-tendon complex (MTC). Nevertheless, active control mechanisms may have limited effectiveness due to intrinsic delays in the reflex pathways and muscle low-pass characteristics. The use of control-oriented mathematical models, aided by analytical methods, help provide insight into neuro-physiology. Control of balance in human upright standing is particularly well suited for modeling, and is also a popular experimental paradigm. This paper examines neuro-physiological basis of postural stability and control in the background of popular biomechanic and neuroscientific models.**

I. INTRODUCTION

HUMAN posture refers to the static disposition of limbs and body parts. Transition among static postures entails movement, which may be categorized as postural, voluntary, skilled, ballistic, phasic, etc. At a broad level goal directed voluntary movements can be distinguished from postural reactions that are initiated in response to internal or external perturbations to the standing posture. While postural movements are usually performed in the range of 5-10 Hz, skilled voluntary movements can be performed at much higher speeds. Static posture or postural equilibrium implies balance of forces and moments acting on the body, that requires the center of gravity of the whole body to be positioned over the base of support (BOS) – the area under the two feet. In the context of voluntary/postural movement, the dynamic equilibrium refers to the balance of forces and moments (including inertial moments) on the limbs and body parts when in motion. The center of mass (COM) in the case of dynamic equilibrium is usually not restricted to the BOS. For example, COM during walking is located outside of the BOS 80% of the time [1].

The neuro-physiological processes involved in postural control and movement regulation include: the central nervous system (CNS), comprising brain and spinal cord; the peripheral nervous system, comprising afferent and efferent pathways; the musculoskeletal system comprising skeleton driven by the muscle-tendon actuators; and, the sensory system composed of a variety of distributed sensory receptors, including muscle spindle (MS), GTO, joint, subcutaneous, somatosensory, and mechanoreceptors. These processes collectively describe the neuro-musculo-skeletal control system (NMSCS) that plans, organizes, executes, and regulates the motor modalities in the body.

CNS control of posture and movement is exerted through a command hierarchy that is believed to have higher, middle and lowest levels [2]. The highest level operates in the association cortex and develops overall motor plans or strategies. The middle level that converts strategy into motor programs, resides in the sensorimotor cortex, the cerebellum, the putamen loop of basal ganglia, and the brain stem. The lowest level operates in the spinal cord; it translates motor programs into muscular activity, which it servo-regulates through stretch reflexes. Although neural mechanisms regulating postural control are unknown, evidence suggests that the hierarchal controller for postural adjustments resides in supraspinal circuits, possibly in the brainstem [3]. Active control of movement trajectory is achieved via continuously varying the motor neurons firing rates that stimulate antagonist muscle pairs. Primary movement stability during trajectory formation is provided by the visco-elasticity inherent to the musculo-skeletal system, and the resulting mechanical stiffness of the muscle-joint structures. Stiffness is maintained at constant levels in static posture, but dynamically varied during performance of skilled voluntary movements. For example, during walking stiffness is maximized at a time in the step cycle when the extensors must support the weight of the body.

The state of the musculoskeletal system may be represented via such variables as muscle length, tone, stiffness, rate of shortening, etc. These variables are monitored by sensory receptors and transmitted via afferent pathways to the CNS, where they are integrated and processed with other proprioceptive information and stimuli (tactile, somatosensory, visual, and vestibular) to generate muscle activation commands. These commands are then sent via efferent pathways to the muscle actuators where they energize the motor units (MUs), each MU comprising of a single motor neuron and the muscle fibers it stimulates. The ensuing contractile action by muscle fibers facilitates movement in support of the intended task.

While mechanisms of postural and voluntary movement are similar in nature, in this discussion we concentrate on the former. Specifically, we examine the neuro-physiological bases of postural stability and control in the background of popular neuro-scientific and biomechanical models.

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II. Postural Adjustments and Stability

A postural response or postural reaction refers to muscle activations that help maintain body alignments of mass centers amid internal and external disturbances. Postural adjustments form an integral part of any motor program. Internal disturbances induced by involuntary and voluntary movements are anticipated by the CNS, which accordingly modulates the background activity of postural muscles. Such modulations are referred to as anticipated postural adjustments, or APAs. For example, in arm raising experiments, lower back and leg muscles were observed to contract 80 msec before the deltoid contraction, aimed at stabilizing the posture for the shoulder flexion task [4]. Postural adjustments involve both proactive and reactive mechanisms, and are either triggered by muscle stretch through long loop responses, or by visual or vestibular input.

Postural stability, i.e., our ability to maintain static posture amid disturbances, constitutes an important attribute of the physiological system. Our stance underlies incessant postural adjustments aimed to counter multidimensional disturbances to the standing posture. These include heartbeat, breathing, lateral adjustments, head-arm-and-trunk (HAT) movements, etc. The overall effect of these disturbances is to displace the body COM from its intended location over the BOS. A time plot of instantaneous COM position in the horizontal plane displays random (Brownian) motion in all directions. Since a majority of the DOFs involved in postural adjustments (ankle, knee, and hip joints) lie in the sagittal plane, the adjustments predominantly affect the COM position in the anterior-posterior (AP) direction. The COM in normal stance lies about 5cm ahead of the ankle joint in the anterior direction thus requiring support. Constant activation of the soleus and/or tibialis muscles in the lower leg generates necessary palantarflexion torque that counters gravity and helps maintain balance. Somatosensory inputs are believed to provide the CNS with an estimate of the instantaneous COM location over the BOS. Nevertheless, COM being primarily a derived quantity, it is debatable whether the CNS actually monitors COM position [5].

A. Determinants of Postural Stability

Center of Mass/Center of Pressure: COM and/or COP position relative to the BOS are often used as measures of postural stability. COM represents the center of gravity of the whole body. The COM position in three-dimensional space (longitudinal, lateral, vertical) can be computed through the application of the principles of mechanics. In postural stability studies the sagittal plane mechanics and the anterior-posterior position of COM are dominant. The COP refers to the projection on the ground plane of the vertical ground reaction force distribution under the BOS. In human motion studies, COP is often obtained from the force plate data. It is well-known that the whole body COM under static conditions and the COP under dynamic conditions should be restricted to the BOS to maintain postural stability.

Extrapolated Center of Mass [6]: In a dynamic situation along with the COM position, the COM velocity also enters in the stability determination. As, even though the COM is located over the BOS, balance may be impossible if the COM velocity is directed outward, and vice versa. A measure of stability that is applicable in dynamic situations has been described as 'the extrapolated COM', or XCOM, that includes a measure of the instantaneous COM velocity. It has been shown that, in order to maintain dynamic stability, the XCOM must be restricted to the BOS. While COM has been observed to keep a distance of at least 5 cm relative to the COP, XCOM and COP approach each other quite closely, around 2.5cm at the instant of foot contact. Further, the XCOM enables reasonable formulation of the stability requirements for walking in the case of and IP model based on a margin of stability that involves XCOM relative to the upper limit of the BOS.

Feasible Stability Region [7]: A person's ability to terminate movement and to restore standing balance during an impending fall is largely based on how that person negotiates various physical constraints arising from physiological, anatomical, and environmental origins. A region of stability in the COM phase plane can be predicted based on the physical constraints of muscle strength, size of BOS, and floor surface contact forces within an environment. Such a region has been shown to be useful for investigation of postural stability. FSR framework has been used to quantitatively describe the postural stability of a motor task performed under a variety of constraints. Researchers further demonstrated that an overlapping FSR existed in the case of the slipping and non-slipping conditions, demonstrating existence of movement strategies that restored stability and balance during an impending fall.

III. Mechanisms of Postural Stability

A. Biological Motor Servo

Postural and movement stability in the body is enabled by the biological servomechanism known as the lower-level motor servo that consists of muscles, stretch receptors and neural pathways leading to and from the CNS. The servo system generates reflexes aimed at relieving muscle tension, and returning muscle tone to its pre-set level. The simplest of these reflexes is the monosynaptic stretch reflex, which is caused by a single synapse from sensory neuron onto the motoneuron inside the spinal cord.

Muscle behavior in vivo has been a topic of great interest. In 1938, A.V. Hill proposed his well-known model of muscle mechanics, [8], which consisted of three components: a contractile element (CE), a series elastic (SE) element, and a parallel elastic (PE) element. With some modifications Hill's model has been followed by a majority of the researchers. The CE element in Hill's model is normally described in terms of muscle length-tension and force-velocity characteristics, while SE and PE elements can be modeled as linear or nonlinear. Muscle behavior being inherently nonlinear, muscle modeling represents the most complex part of physiological modeling. However, simple second order models are considered adequate over operating range.

The presence of SE element in Hill's model gives rise to the spring-like behavior of the muscle that is analogous to stiffness in the linear case [9]. The spring-like behavior reflects both intrinsic MTC stiffness and the added stiffness due to spinal reflexes. In the absence of proprioceptive feedback spring-like behavior is dominant; however, spring-like behavior is retained in the presence of proprioceptive feedback, playing an important role in the maintenance of posture and in the formation of movement trajectories. Specifically, spring-like behavior causes the joint trajectories to exhibit stable equilibrium behavior in the absence of sensory feedback.

Muscle spindle (MS) is a small stretch sensitive sensory organ, which contains intrafusal fibers with both sensory and motor innervations, and lies parallel to the extrafusal muscle fibers. The sensory endings at rest maintain a certain firing rate into group Ia and group II afferents resulting in a preset muscle tone. The intrafusal fibers attached to the muscle spindle are energized by the fusimotor neurons (or γ-motoneurons). The extrafusal and the intrafusal fibers are maintained at the same length causing the spindle to act as a regulator of muscle length. Muscle spindles are known to exhibit prominent nonlinearities. The sensory fibers in the spindle are sensitive to small length changes of the order of 1μm. The velocity sensitivity of the spindle is initially high and decreases at higher velocities.

B. *Active Control vs. Passive Stiffness*

Postural stabilization entails both active and passive mechanisms at muscle and spinal level as well as visual and vestibular processes. Active torques results from muscle contraction in response to CNS commands from higher centers and/or reflex loops, whereas, passive torques results from the intrinsic stiffness and viscosity in the muscle and surrounding tissue such as ligaments and tendons.

In an effort to characterize empirical data obtained from translating support surface experiments, some researchers (e.g., [10]), have proposed stiffness-only models of postural control lacking significant active or reactive components, except for background setting of the stiffness parameters. Other researchers, [11], however, noted that the lowest ankle stiffness to support a stiffness-only model was 1835 Nm/rad, which was considerably higher than the direct measurements of ankle stiffness that showed a range of 250-400 Nm/rad with a large bias torque of 100 Nm [12]. Thus, ankle stiffness alone was insufficient, and stability augmentation through direct CNS involvement was required to stabilize body sway. Nevertheless, theoretical considerations involving intrinsic delays in the reflex pathways and the low-pass characteristics of the muscle response tended to discount any reflex nature of the stabilization mechanisms.

Having excluded a dominant effect of muscle stiffness as well as reflex-dominated reactive stabilization, the role played by the CNS in active control of stance remains an open an intriguing question. Researchers, [11], believe that the central computational processes carry out two main functions: 1) integrating the multisensory information into unifying estimate of the state vector and 2) compensating the transmission delays with an anticipatory action, i.e., a short-time prediction of the postural time series. While the exact implementation of these functions in the CNS is unknown, these functions seem to be necessary for postural stability.

IV. Models in Postural Control

A. *Internal Models*

Internal models that mimic the behavior of the natural processes represent an important theoretical concept in motor control. Such models are widely used by neuroscientists to explain the internal working of the neuro-musculoskeletal system. It has been proposed that the CNS internally simulates the dynamic behavior of the motor system in planning, control, and learning of movement. Internal models can be categorized as forward (predictor) models or inverse (controller) models. According to Wolpert et al., [13], forward models may be useful in solving four fundamental problems in computational motor control: 1) to overcome the transmission delays in the motor actions associated with feedback control; 2) to anticipate and cancel the sensory effects of movement; 3) to use the error signal between desired and actual sensory outcome of a movement for motor learning; and, 4) for state estimation leading to re-afferent sensory correction. Although shown to be of theoretical use, the existence of internal forward models in the CNS is still a topic of debate.

B. *Inverted-Pendulum (IP) Models*

The IP model of the body has been widely used by researchers for analyzing postural sway stabilization. It includes a simplified representation of the musculo-skeletal system that comprises a single rigid segment rotating around the ankle joint, and acted upon by the force of gravity and the net torque produced by the MTCs surrounding ankle joint. An essential requirement for local asymptotic stability in this model is that the sum of the (positive, stabilizing) ankle-joint stiffness and the (negative, destabilizing) gravitational stiffness must be larger than zero, i.e., $K_{MTC,eff} + K_g > 0$, where K_{MTC} represents the stiffness of the MTC and K_g represents gravitational stiffness [14]. Experimental data from various studies supports the use of an IP model of postural control. For example, researchers who studied static-dynamic stimulus combinations and response asymmetries among normal subjects and vestibular loss patients concluded that the IP simplification was legitimate, and that human upright stance could be modeled in terms of continuous multi-sensory feedback control [15].

In the context of IP model, postural sway may be interpreted as the result of noise that acts on a system that has an equilibrium that is locally asymptotically stable. Further, the interaction of the combined active and reactive stiffness may be used to identify the minimum conditions needed for quiet standing. An IP model with moving feet was successfully used to investigate slipping, sliding, and falling behavior. Other researchers used system identification methods to separate intrinsic and reflexive components of the applied torques in the human arm during posture maintenance task. They concluded that feedback gains varied considerably with the frequency content of the disturbance signal. In particular, substantial reflexive dynamics were observed for low-frequency (<3 Hz) input signals and for near-sinusoidal inputs (>1.5 Hz) [16].

C. *Multi-Segment Models*

Though simple IP models have been shown to be effective in analyzing postural sway, better approximation of postural dynamics can be obtained by adding ankle and hip degrees-of-freedom (DOFs) to the model. In postural perturbation experiments, the body has been observed to move as a multi-link structure with extensive knee movements, implying that standing balance depends on the effective control of the torques at the ankle, knee, and hip joints. Researchers (e.g., [17]) believe that the single-link inverted pendulum model provides a less conservative estimate of minimum stiffness, i.e., more stiffness is required at each joint to preserve stability when rotation is permitted at the knee and hip joints. Further, the interaction of the combined active and passive stiffnesses at the ankle, knee, and hip might be used to identify the minimum stability conditions for quiet standing.

V. Controllers for Postural Movements

A. *Proportional-Integral-Derivative Controllers*

In model-based studies, the decision-making function of the CNS is commonly represented by a control mechanism such as proportional-integral-derivative (PID) control. In movement science, researchers have often used an IP model coupled to a simple spring-damper servomechanism that naturally develops into PID control action. Further, it has been reported that experimentally observed gain and phase data in human postural control fits an IP model with a PID controller [18]. Thus PID control serves as popular modeling analogue of a more complex neural controller. Recent studies have proposed a high gain PD (proportional-derivative) controller as replacement of PID control [19]. However, computer simulation studies with neurophysiological controller models have shown that compared to PD, a PID controller better symbolizes the CNS control action [20].

B. *Optimal Control of Postural Movements*

Robust and optimal design techniques that combine estimation and control theory, whereby partial state measurements are used to construct the state variables for a linear feedback controller, have been successfully applied to biomechanical models. These methods provide stability and performance robustness against modeling errors, which limit high-performance control systems design. The robust and optimal control design aims for maximum postural stability amid unmodeled dynamics, noisy measurements, neuro-muscular deficits, and external perturbations. A major step in the application of robust and optimal design methods to a biomechanical model involves formulation of the postural control problem in the robust control framework, followed by controller synthesis using, e.g., μ–synthesis commands. Alternatively, model predictive control (MPC) has been effectively used to generate stable walking motions [21].

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