CHAPTER 31

Human standing posture: multi-joint movement strategies based on biomechanical constraints

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We developed a theoretical framework for studying coordination strategies in standing posture. The framework consists of a musculoskeletal model of the human lower extremity in the sagittal plane and a technique to visualize, geometrically, how constraints internal and external to the body affect movement. The set of all feasible accelerations (i.e., the "feasible acceleration set" or FAS) that muscles can induce at positions near upright were calculated. We found that musculoskeletal mechanics dictate that independent control of joints is relatively difficult to achieve. When muscle activations are constrained so the knees stay straight, to approximate the typical postural response to perturbation, the corresponding subset of the feasible acceleration set greatly favors a combination of ankle and hip movement in the ratio 1 : 3 (called the "hip strategy"). Independent control of these two joints remains difficult to achieve. When near the boundary of instability, the orientation and shape of this subset show that the movement strategy necessary to maintain stability, without taking a step, is quite restricted. Hypothesizing that regulation of center-of-mass position is crucial to maintaining balance, we examined the feasible set of center-of-mass accelerations. When the knees must be kept straight, the acceleration of the center of mass is severely limited vertically, but not horizontally. We also found that the "ankle strategy", involving rotation about the ankles only, requires more muscle activation than the "hip strategy" for a given amount of horizontal acceleration. Our model therefore predicts that the hip strategy is most effective at controlling the center of mass with minimal muscle activation ("neural effort").

Key words: Standing; Posture; Coordination strategies; Dynamics; Motor control

Introduction

Considerable success has been achieved in devising and performing experiments to study sensorimotor control of standing posture. One reason is that motor control output variables (e.g., body segment kinematics, ground reaction force and EMG signals) can be measured with precision. Perhaps more importantly, ingenious methods to manipulate the inputs to this sensorimotor system have been devised and implemented. For example, the surface on which subjects stand can be moved (Nashner, 1976, 1977), and vision can be blocked, using blindfolds, or altered by dynamically moving the visual surround (Nashner et al., 1982; Black et al., 1988). Additional insight into sensorimotor control has been achieved by observing how patients with neurological disorders (e.g., vestibular loss, or dysfunction) react to these controlled disturbances (Allum and Pfaltz, 1985; Allum et al., 1988; Horak et al., 1990).

However, theoretical constructs to analyze experimentally based hypotheses are few, even though theories seem to offer much potential (Nashner and McCollum, 1985; Nashner et al., 1989; Gordon, 1990). The complexity of the biomechanics of the system being controlled is probably what has stymied theoretical investigations. For example, movement about a single joint is complex enough to analyze (Gottlieb et al., 1990), much less the combined properties of many muscles acting in concert
on a multi-segment body (Zajac and Gordon, 1989). Nevertheless, we feel that theoretical studies of posture are a useful adjunct to experimental studies, and perhaps essential to understanding postural control of this complex biomechanical system.

Many experiments in posture have relied on rotating or translating support surfaces to produce perturbations. Researchers have reported consistent strategies in response to such perturbations, and termed these responses “ankle” and “hip strategies” (Nashner and McCollum, 1985; Horak and Nashner, 1986) referring to the joint about which most of the movement occurred. They reported that responses to small disturbances typically entailed the ankle strategy, with a combination of ankle and hip strategies being chosen for shortened support surfaces or faster disturbances (Diener et al., 1988).

Nashner and McCollum (1985) also established a theoretical basis for analyzing posture through examination of the equations of motion for a three-segment model of the human body in the sagittal plane. They established a coordinate system for the body position space using ankle, hip and vertical axes. Since postural movements occur about the ankle and hip (Nashner, 1977), the vertical axis can be disregarded. In the ankle-hip plane, they sketched vectors corresponding to the accelerations produced by respective activation of ankle, thigh and trunk muscles. They suggested that movement strategies are chosen so as to minimize the number of muscles activated, and predicted corresponding acceleration trajectories.

Nashner et al. (1989) explored the constraints imposed by mechanics and sensory thresholds, delineating frequencies and amplitudes for which movements can be detected by sensors (and therefore controlled) and for which the body is capable of producing driving torques. This work was instrumental in delineating the frequencies and amplitude ranges within which postural movements may occur.

While research to date has been successful in quantifying postural behavior, little work has been done to illustrate what causes this behavior to be produced. We believe that postural movement is shaped by the organization of the mechanical characteristics of the human body, as well as the nervous system which controls it. To this end, an analysis of the biomechanics of standing posture will help us understand control strategies underlying postural behavior.

The biomechanics of the musculoskeletal system are complex. Zajac and Gordon (1989) showed that muscle’s contribution to acceleration depends not only on the dynamics of the body segments, but also on other characteristics, including the maximum achievable isometric force, moment arm and the length of muscle fibers. Gordon (1990) explored the vectorial contribution of muscles to the acceleration of joints, and proposed computing the bounds on acceleration as a means to visualize and study synergies and constraints on motion.

In this paper, we assemble a musculoskeletal model for the computation of the bounds on acceleration, which we term the “feasible acceleration set” (FAS), and examine the mechanics of the ankle and hip strategies in relation to various constraints. We also explore the hypothesis that acceleration of the body center of mass while minimizing net muscle activation is a central objective in maintaining posture. In so doing, we provide partial explanation of control strategies to counteract perturbations to the upright position.

**Analysis**

**Musculoskeletal model**

The model of the lower extremity incorporates the dynamics of both the body and musculotendon actuators (muscle in series with tendon). The body, modeled as a four-segment linkage comprised of the foot, shank, thigh and head-arms-trunk (referred to as the “trunk” hereafter), was assumed to move in the sagittal plane only (Fig. 1). The segment lengths and inertial parameters were prescribed for an “average” adult male (Pandy et al., 1990). The ankle, knee and hip joints were modeled as simple hinge-joints in the sagittal plane. However, a more complex kinematic model was used to characterize the moment arms about the knee (Delp et al., 1990). The path of each musculotendon actuator was de-
We are confident that this type of model is adequate for simulating human movement. A similar musculoskeletal model incorporating additional velocity-dependent terms has been used to produce simulated jumps that reproduce the salient kinetic, kinematic and muscle coordination features observed in humans jumping to their maximum achievable heights (Pandy et al., 1990; Pandy and Zajac, 1991).

Specific to this study of standing posture, we assumed the motion to be quasi-static, i.e., the velocity terms in the dynamical equations, as well as muscle fiber velocities, to be small. In addition, muscle activation dynamics were assumed to be fast (instantaneous) in relation to the movement. Our analysis concerns only selection of acceleration vectors (the set of joint angular accelerations describing movement) at a given instant in time, and neglects muscle activation sequencing. For simplicity, the muscles within each of the 14 muscle groups were assumed to share common activation commands from the nervous system. We also restricted study to configurations in which the feet are flat on the ground, limiting our study to postural movement without taking a step. This constraint reduces the number of degrees of freedom in the model to three.

The feasible acceleration set

The equations of motion determining the joint angular acceleration vector $\ddot{\theta}$ (for ankle, knee, and hip joint angular accelerations) can be written as:

$$\ddot{\theta} = M^{-1} (R \cdot F_0 \cdot F_t \cdot a + G) \tag{1}$$

where $a$ is the vector of normalized muscular activation levels, neglecting whether activation is achieved by frequency modulation or recruitment (Stein, 1974); $M$ is the mass matrix relating torques to angular accelerations; $R$ is the moment arm matrix; $G$ is the vector containing gravity terms; and $F_t$ and $F_0$ are diagonal matrices characterizing muscle's force-length and peak isometric force characteristics, respectively (Zajac, 1989).

Normalized activation level for each muscle $i$ is confined by the inequality $0 \leq a_i \leq 1$, thus con-
fining feasible activations to an \( m \)-cube (for \( m = 14 \) muscle groups). Eqn. 1 is used to compute the corresponding set of inequalities on the joint angular accelerations \( \theta \), transforming the \( m \)-cube into a polyhedron in the three-dimensional space defined by ankle, knee and hip axes (Gordon, 1990). The surface of this polyhedron is the outer bound on all possible accelerations, and we denote its entire volume as the theoretically feasible acceleration set (FAS). The FAS represents all possible body accelerations for a given quasi-static configuration.

A FAS can be analyzed to assess “ease in movement” in a specific direction subject to the constraints applicable to that particular FAS. Because any desired acceleration vector from the origin can be viewed in relation to an acceleration vector that is in the same direction but extended to intersect the FAS boundary, the desired acceleration vector can be expressed as a percentage of the maximum feasible acceleration in that direction. If this percentage is low, then this acceleration intensity can be achieved with little muscle activation or “neural effort”, so that the “ease of movement” is high.

Equivalently, if maximal neural effort (muscle activation) produces accelerations reaching the boundary of the feasible acceleration set (the surface of the polyhedron), then correspondingly smaller efforts can be used to produce accelerations reaching the boundary of a scaled-down feasible acceleration set. In this sense, accelerating at a specified magnitude in directions for which the FAS boundary is a large distance from the origin requires less muscle activation (less neural effort) than that to accelerate at the same intensity in other directions. Note that the nervous system need not employ such a scaling method for choosing muscle activations; it may in fact choose other patterns for achieving any submaximal acceleration. However, experimental evidence (Dienert et al., 1988) has shown that postural responses, including muscle activation, do tend to scale with platform stimulus parameters.

Because activation level is normalized with respect to a muscle’s peak isometric force, which in turn is computed from the physiological cross-section area (Pandy et al., 1990), the index of neural effort is equivalent to an index of muscle stress (force divided by cross-section area).

We devised a method for computing and displaying the acceleration bounds describing the FAS polyhedron in ankle-knee-hip acceleration space (the ankle-knee-hip FAS). Mathematically, this involves finding the convex hull of the affine transformation of an \( m \)-cube for the \( m \) muscle groups. Additional biomechanical constraints are accounted for by displaying the FAS with respect to corresponding half-spaces (described by planes) defined by these constraints. The intersection of these half-spaces shows how these constraints interact to restrict the repertoire of possible movements.

In their analysis, Nashner and McCollum (1985) emphasized that responses to postural disturbances involve ankle and hip, but little knee rotation. The consequence of avoiding knee rotation is to limit the ankle-knee-hip FAS to that region which intersects the plane defined by zero angular acceleration of the knee. This intersection, called the ankle-hip FAS, is bounded by a polygon, because the constraint reduces the three-degree-of-freedom FAS to two degrees-of-freedom. Note that this constraint acts on the controls, limiting the ankle-knee-hip FAS to those combinations of muscles that would keep the knee straight (\( \theta_{knee} = 0 \)), as opposed to a constraint on the positions, which would be analogous to an external brace on the knee and would require different equations of motion.

We also studied how the constraint of keeping the feet flat on the ground restricts the ankle-hip FAS. This “flat-foot constraint” is actually defined by two constraints, corresponding to accelerations that cause lifting of the heels and of the toes off the ground, respectively. Thus, the feasible accelerations satisfying these two constraints are a subset of the ankle-hip FAS, and are designated the flat-footed ankle-hip FAS.

We also determined the effect of body segment orientation on the ankle-hip FAS. The matrices \( M \), \( R \) and \( F_p \) as well as the vector \( G \) of Eqn. 1, are dependent on body position. Thus, the FAS must also depend on body configuration. This dependence
may dictate different coordination strategies in response to perturbation at different body positions.

Because horizontal center-of-mass position is the final determinant of stable balance and because long-duration upright standing posture seems to demand little effort, we hypothesized that one of the objectives of standing posture is to control the center-of-mass horizontally with minimal neural effort (i.e., muscle activation). Accelerations of the ankle, knee and hip define corresponding accelerations of the center of mass horizontally and vertically. Thus, feasible ankle-knee-hip accelerations define corresponding feasible accelerations of the center-of-mass (i.e., center-of-mass ankle-knee-hip FAS). To examine the hypothesis stated above, we calculated the center-of-mass ankle-knee-hip FAS and the center-of-mass ankle-hip FAS (i.e., knees kept straight). The toe- and heel-off constraints from ankle-hip space were likewise transplanted to the space of vertical and horizontal center-of-mass accelerations.

Results

We computed the ankle-knee-hip FAS for a near-upright standing position, with the knees unlocked, and displayed it as a polyhedron with solid lines separating its faces (Fig. 2.4). The polyhedron is approximately 16 times longer than it is wide, and 33 times wider than it is thick. Its long axis corresponds to a combination of simultaneous ankle, knee and hip angular acceleration in the ratio of 1 : 2.3 : 1.5, with magnitudes of approximately 1500 rad/s² in extension and 830 rad/s² in flexion. Its width axis corresponds to a combination of simultaneous ankle, knee and hip acceleration in the ratio 1.6 : 1 : −2.7, with magnitudes of approximately 63 rad/s² in extension and 75 rad/s² in flexion (note that the hip component of the ratio is negative, indicating that when the other joints are accelerating into extension, the hip is accelerating into flexion, and vice versa). Finally, the axis pointing in the thickness direction of the polyhedron corresponds to a combination of simultaneous ankle, knee and hip acceleration in the ratio 2.8 : −1.9 : 1, with magnitudes of approximately 2.2 rad/s² in extension, and 2.1 rad/s² in flexion.

To examine the effect of keeping the knees straight on the ankle-knee-hip FAS, the polyhedron is intersected with a plane defined by \( \dot{\theta}_{knee} = 0 \). This intersection is the set of all achievable accelerations when muscle activations are constrained so as to produce no motion about the knee. This ankle-hip FAS, for the same near-upright posture as in Fig. 2, is a polygon (Fig. 3) and is a single slice of the original volume of the ankle-knee-hip FAS. It also indicates the general shape of the cross-section of the polyhedron in Fig. 2, as the \( \dot{\theta}_{knee} = 0 \) plane also cuts through the ankle-knee-hip FAS, though at an
A change in body position affects the FAS because segmental configuration, segmental orientation with respect to gravity, musculotendon lengths, and moment arms, all of which interact to define the FAS (see Eqn. 1), depend on body position. We found, however, that the ankle-hip FASs are similar in shape and orientation for positions near upright standing (Fig. 4 shows ankle-hip FASs for 12 separate positions. The position boundaries within which the body can maintain stable balance, with the center-of-mass located above the foot between the toes and heels, is shown by the dashed lines).

Fig. 3. Ankle-hip FAS for near-upright posture, specifying set of achievable accelerations while keeping the knees straight. This FAS appears in the ankle-hip plane because knee angular acceleration is zero (i.e., the knees are constrained to be straight). Boundaries beyond which the toes or heels lift off the ground are shown as dashed lines. The shaded region (flat-footed ankle-hip FAS) shows the feasible accelerations such that the feet are kept flat on the ground while the knees are kept straight. Accelerations outside this region, but interior to the ankle-hip FAS polygon, are possible but cause either the toes or the heels to lift off the ground. Note that the ankle acceleration scale is expanded.

angle of approximately 52°. The long axis of the polygon corresponds to simultaneous ankle and hip acceleration in the ratio 1 : -3.2 with magnitudes of approximately 72 rad/s^2 in flexion and 62 rad/s^2 in extension. The constraint for keeping the feet flat on the ground is manifested by two lines, as indicated, beyond which either the toes or the heels lift off the ground. Thus, the region inside both the ankle-hip FAS and within these constraints represents potential acceleration vectors which keep the feet flat on the ground. This flat-footed ankle-hip FAS corresponds to simultaneous ankle and hip acceleration in the ratio 1 : -3.2, with magnitudes of approximately 67 rad/s^2 in flexion and 43 rad/s^2 in extension.

Fig. 4. Ankle-hip FAS plotted for 12 separate body positions (origin indicating body position for each FAS plot is shown by crosshairs). Body position is indicated by ankle angle and hip angle (as defined in Fig. 1). Dashed lines indicate position boundaries beyond which stable balance cannot be achieved. Left line is boundary for falling forward, right line for falling backward. Note that the FASs are similar in size and orientation throughout the body position space. Inset shows scale of axes for FAS plots.
Therefore body position has a relatively small effect on the FAS and conclusions drawn for the near-upright stance studied above can be generalized to the range of typical postural body positions.

When the body is leaning forward (left-most boundary, Fig. 4), the heel- and toe-off constraint lines (as seen in Fig. 3, but not shown in the FAS plots of Fig. 4) both shift to the left. Similarly, when the body is leaning backwards, the constraint lines move to the right. Thus, the feasible accelerations that move the body toward upright posture (i.e., acceleration vectors pointing away from the boundary in Fig. 4), yet keep the feet flat on the ground, become quite restricted. In fact, if the body should be leaning as far forward or backwards as possible (i.e., the body lies on one of the boundaries, Fig. 4), use of a pure ankle strategy will violate one of the constraints, causing either the heels or toes to be lifted off the ground. We also found that the heel- and toe-off constraint lines both move in toward the origin as the support surface shortens (Horak and Nashner, 1986). In fact, with an extremely narrow surface, we predict that balance in the face of a disturbance cannot be maintained using a pure ankle strategy. Therefore, should the body be near the boundary of stable posture, or should the support surface be narrow, both the hips and ankles must accelerate with quite restrictive relative amounts (i.e., ankle to hip acceleration of about 1 : -3.2).

The center-of-mass ankle-knee-hip FAS (i.e., the set of all feasible accelerations of the center-of-mass derived from moving the ankle, knee and hip) is defined by a two-dimensional polygon because the center-of-mass has only two degrees of freedom, horizontal and vertical, in the sagittal plane (Fig. 5A). The center-of-mass ankle-hip FAS (i.e., the center-of-mass FAS when the knees are kept straight) is a small subset of the entire center-of-mass FAS, and is greatly restricted in the vertical direction (darkened region, Fig. 5A).

Based on the definition of the ankle strategy as acceleration about the ankles only, and the hip strategy as a combination of ankle extension and hip flexion or vice versa, we constructed acceleration vectors corresponding to these strategies. We found that both ankle and hip strategies are effective at ac-
celerating the center-of-mass horizontally, but that the hip strategy requires less neural effort for a given magnitude of horizontal acceleration (Fig. 5B). The toe- and heel-off constraint lines (dashed lines in Fig. 5B) are nearly parallel to the long axis of the center-of-mass ankle-hip FAS.

Discussion

The shape of the ankle-knee-hip FAS indicates that certain combinations of joint accelerations require much less neural effort than others (Fig. 2), as indicated by the distance of the FAS boundary from the origin. At near-upright position, the body mass matrix is relatively close to rank 1, meaning that it is difficult to move the joints independently. However, certain combinations of simultaneous ankle, knee and hip acceleration in extension or flexion (see Results), corresponding to movement along the long axis of the polyhedron, require relatively little effort. Movement combinations in the direction of the width or thickness of the polyhedron require much more effort. Thus, the biomechanical relationships of the human body place severe constraints on the relative effectiveness of and neural effort associated with movement in different directions.

The ankle-hip FAS (Fig. 3) shows that keeping the knees straight greatly limits the set of achievable accelerations. It is important to note that keeping the knees fixed is not equivalent to keeping the knee muscles inactivated, as knee torque is still needed to maintain the constraint. Still, independent motion about the ankle and hip joints is constrained by biomechanics such that many acceleration vectors require a great deal of neural effort, or are limited in feasible magnitude. Combinations of ankle and hip extension in the approximate ratio 1 : −3.2 are easily achievable. However, acceleration about the ankles only or the hip only is relatively more difficult to achieve.

Horak and Nashner (1986) defined the “ankle” and “hip” strategies based on a combination of electromyogram, force plate and kinematic patterns. Because our model does not provide electromyograms or horizontal shear forces, we define similar strategies based purely on observation of joint accelerations, for purposes of comparison. Our own ankle strategy is defined as acceleration about the ankle joint only, keeping the knee and hip fixed, and would correspond to movement along the ankle axis of Fig. 3. Our hip strategy corresponds to acceleration of the ankle and hip in opposite directions, with the hip accelerating about three times more than the ankle, approximately along the toe- and heel-off constraint lines shown for the ankle-hip FAS (Fig. 3). These definitions help to explain our findings in comparison with the work of Horak and Nashner (1986). Such a hip strategy is fairly well-suited to keeping the feet flat on the ground, while requiring less neural effort for a given amount of acceleration, due to the length of the ankle-hip FAS in this direction.

A disturbance tends to accelerate the body away from the upright position, requiring muscles to arrest the motion and bring the body back to erect stance. Faster disturbances require larger acceleration vectors, and because large ankle accelerations are likely to result in toe- or heel-off, the hip strategy is more effective at countering large disturbances and accelerating the center-of-mass toward a more stable position while maintaining the feet flat on the ground.

At the extremes of the stable flat-footed posture (positions near the boundaries shown in Fig. 4), employment of the ankle strategy (i.e., acceleration along the ankle axis toward upright posture) to restore the body to the upright position will easily violate the flat-feet constraint (see Results). The only recourse is to use the hip strategy to bring the body away from the boundaries before using the ankle strategy to restore the nominal position.

Similarly, shortened support surfaces will also require greater reliance on the hip strategy. The toe- and heel-off constraint lines both move closer to the origin with shorter support surfaces, further limiting the efficacy of the ankle strategy.

In our analysis of the center-of-mass FAS (Fig. 5A), we found that for motions in which the vertical acceleration of the center-of-mass is limited
(\dot{\gamma} \approx 0), the ankle-hip FAS allows for nearly the entire range of horizontal acceleration of the center-of-mass. Thus, preventing knee motion has a more dramatic effect in limiting vertical rather than horizontal acceleration. Despite the limiting effect of the knee constraint on the set of achievable accelerations (see Results), its effect on accelerating the center-of-mass horizontally is negligible.

In accelerating the center-of-mass, the ankle strategy requires more neural effort than the hip strategy, which accelerates the body approximately along the toe- and heel-off constraint lines (Fig. 5B). Note also that in comparison to the ankle strategy, the hip strategy can accelerate the center-of-mass horizontally faster and still keep the feet flat on the ground.

A principle favoring acceleration of the center-of-mass while minimizing neural effort to select movement strategies would therefore predict predominant use of the so-called hip strategy, especially when the feet are to remain flat on the ground. Allum et al. (1989) have reported a multi-link strategy similar to our hip strategy in response to relatively fast perturbations. However, Horak and Nashner (1986) have shown that the preferred strategy involves rotation primarily about the ankles in response to relatively slower perturbations, although their experimentally observed ankle strategy may not exclude knee and hip motion entirely, as the theoretical ankle strategy does here. They have also reported use of a hip strategy (similar to the hip strategy defined here) in the case of shortened support surfaces. Since a shortened support surface brings the toe- and heel-off constraint lines closer to the origin (Fig. 3), we would expect greater reliance on the hip strategy.

There are a number of possible reasons for favoring the ankle strategy over the hip strategy in response to slower perturbations. For example, acceleration of the center-of-mass often tends to move the body away from the erect posture, so maintaining balance (i.e., controlling center-of-mass position) is sometimes contrary to the objective of maintaining upright stance. In addition, motion about the hips also causes relatively large movement of the head, and thus may affect vestibular sensors within the skull, which may have implications on sensory feedback to the nervous system.

We can unify the ankle and hip strategies into a single set of criteria for choosing the acceleration vector to restore upright posture, as follows. The speed and position of perturbation specify a minimum response acceleration to keep the body from exceeding the position constraints. The ankle strategy appears to be employed to the extent that the toe- and heel-off constraints allow (as determined by support surface length). If the corresponding center-of-mass acceleration cannot sufficiently counter the perturbation, further acceleration is achieved by moving along the constraint line using a combination of ankle and hip strategies. Mathematically, this can be expressed as a single optimization problem (Kuo and Zajac, 1991).

Our current analysis examines feasible accelerations as determined by body position, neglecting intersegmental velocities. In addition, we do not model the muscle force-velocity relationship (Zajac, 1989). Preliminary investigation has shown that such velocity effects are relatively small, but further work is necessary to quantify their actual contribution. Variations in musculoskeletal model parameters will also cause changes in the FAS. However, we have found the general shape and orientation of the FAS to be relatively insensitive to variations, thus justifying the qualitative results reported here.

Additional work will elucidate the effect of employing ankle and hip strategies on movement and rotation of the head. The effect these strategies have on the sensory input to vestibular sensors may have an impact on which strategies are preferred. Such work may also bring greater understanding to the mechanisms of head rotation during standing.

Finally, we hope that further analysis will provide insight into how to design simple, implantable controllers for stimulating paralyzed muscles to restore upright standing to paraplegics. If control laws can be mapped into the strategies, as defined here,
perhaps a fast feedback algorithm can be designed for multi-joint postural control of paraplegic standing.

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References


