INFLUENCE OF EXTERNAL FORCES ON THE CONTROL AND PERFORMANCE OF A MINIMUM TIME SHOULDER FLEXION TASK

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INTRODUCTION

Despite a near continuous presence in space for the past 30 years, very little is understood about how humans adapt their motor control processes to the novel environment of microgravity. Opportunities to investigate how humans adapt their motor behavior to weightlessness have been fairly limited. Although several studies have been conducted during space flight, which have made use of a video-based motion system to record movement, these studies have only involved one or two subjects (Massion et al., 1993; Roll et al.,1993; Clement et al., 1984). Furthermore, collecting human movement data during space flight is extremely costly in terms of time and financial resources. One supplement to these costly investigations is the use of dynamic computational models of the human body which utilize optimization routines to determine optimal (skillful) movement patterns. Removing gravity from the model's landscape enables investigators to determine the necessary control adaptations to weightlessness during a particular movement.

Previous work in our laboratory has focused on understanding how weightlessness impacts the ability of subjects to perform rapid arm raises during a variety of experimental conditions. In particular, we were interested in how the removal and subsequent restoration of the foot support surface alters the optimal segmental coordination used to perform the movement in microgravity. In addition to our ongoing space flight experiments, we addressed this issue through the development of a 3 segment dynamic computational model which employed optimization routines. Our results indicate that optimal segmental coordination needs to be modified during movement in weightlessness and therefore suggest that humans must learn new patterns of coordination to accomplish their movement objectives.

METHODS

We used a simple mathematical model of the human body to simulate the bilateral arm raise task, and obtained three minimal-movement-time solutions: a) with normal gravitational force (1G) while on the ground, b) with 0G while free floating, and c) with 0G but with the feet attached to the support surface. These environmental conditions were identical to the ones experienced by our subjects during the space flight experiments. Three rigid segments represented the entire body and movement was allowed only in the sagittal plane. The three segments were: 1) the feet, 2) the body (head, torso and legs) and 3) the arms. There were two revolute joints, one between the feet and body segments, which simulated the ankle joint, and one between the body and arms, which simulated the shoulder joint. The feet-to-ground interaction was modeled with two two-dimensional springs, positioned one at the heel and one at the toes, which allowed the feet to either leave or slide along the ground (Anderson, 1995). All conditions used identical anthropometric and strength parameters (Winter, 1990) except for the mass and inertia of the feet in the 0G attached condition. There, the feet's mass and inertia were set to those of the Space Shuttle.

Kane's method of describing dynamical systems was used to derive the equations of motion (EOM). The software package Autolev (OnLine Dynamics Inc.) was used to derive the model's EOMs, and to produce C code for the forward simulation. The simulation code was modified to accept joint torque values in the parameterized form (Pandy et al., 1992) of eight control nodes for each joint and to fit our optimization routines. Linear interpolation was used within nodes for the small time steps required by the integrator.

In general, the equations of motion governing our model's motion have the following form: $\{A(q)\}\ddot{q} + \overline{C}(q,\dot{q}) + \overline{G}(q) + \overline{T} = 0$

Where: $\{A(q)\}$ is the mass matrix; q is the generalized coordinates vector; $\overline{C}(q,q)$ is the Coriolis and centrifugal forces vector $\overline{G}(q)$ is the gravitational forces vector; $\overline{T}$ is the joint torque vector.

The mass matrix is always symmetric and therefore invertible. Thus, the above equations can be solved for the accelerations of the generalized coordinates, used to describe the model.

To limit the range of motion of the free-floating model's ankle joint, within physiological levels, we modeled the ankle joint ligaments with two rotational springs and dampers. Those springs were very similar to the vertical feet-to-ground springs and the following equation describes their torque magnitude: $T_i = 0.5336 \cdot e^{-1150\,(\theta_i - \phi_{0i})} - 10^{3} \cdot \omega_i \cdot \zeta(\theta_i)$

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Where: $\zeta(0)$ is the damping force, given by: $\zeta(\theta_i) = 1/(1 + 10 \cdot e^{300(\theta_i - \zeta_{0i})})$ ϕ_{0i} is the "zero-angle" of the i^{th} ligament; θ_i, ω_i are the angular position and velocity of the i^{th} ligament; $\zeta 0_i$ is a constant which causes a steep rise of the damping torque to occur within 0.05 radians (about 3 degrees) of the maximum range of motion of the joint.

The performance criterion to be minimized was time ($J = t$) to complete the task. The movement was defined by the segments' initial state: anatomical position with no velocity, and final state: segment angles for the 1G and 0G attached, but joint angles for the 0G free (described as the optimal solution is equality constraints: Ψ). A set of inequality constraints (Θ) was used to convey the joint torque maximum limits to the optimum control algorithm. For the solution of these parameter optimization problems we used a sequential quadratic programming routine which required first order derivatives of the performance and constraints. Because of the complexity of the EOMs, we computed all derivatives numerically.

RESULTS AND DISCUSSION

Comparison of the three solutions reveals the effect of gravity and ground reaction force on the dynamics and optimum control of the movement (see figures 1, 2). The 0G models took less time to complete the task, with the free floating model able to perform slightly better than the attached one. These results indicate that the greater the external forces the poorer the performance.

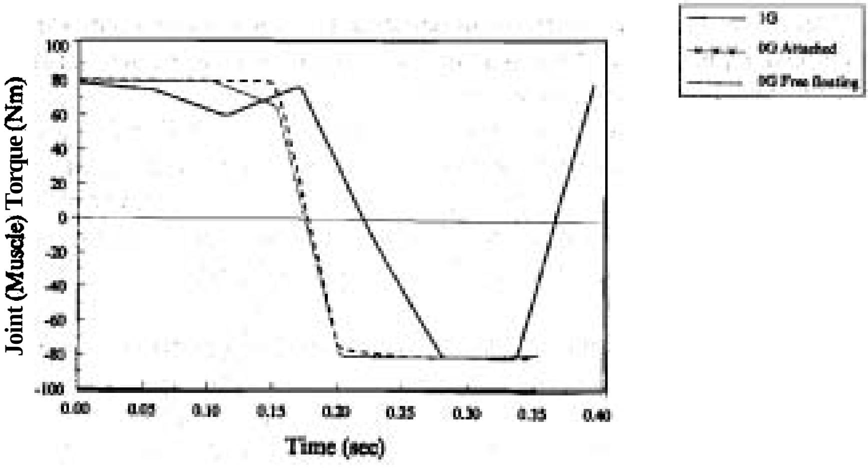

Figure 1. Shoulder torque. Minimal differences between the 0G conditions. The 1G condition was significantly different especially during the later part of the braking phase.

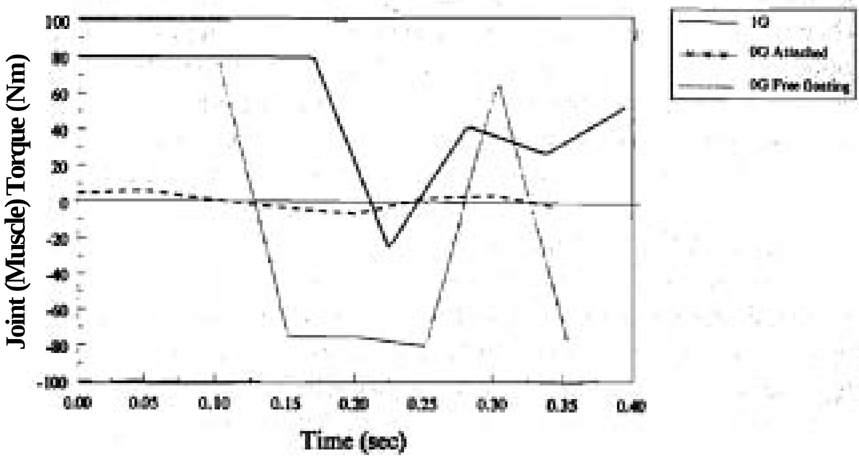

Figure 2. Ankle torque. Varied across all conditions. Notice the limited dorsiflexion torque of the 1G model.

When comparing the three models, it is important to remember that successful completion of the task involves the final state of the arms, body and feet. Because of the nature of the task, the 0G free floating position had slightly different segment state requirements. This could account for the slight performance difference, but should have had no effect on the coordination. The minimal differences between the two 0G conditions in performance, shoulder joint torque and arms kinematics indicate that the large differences in the ankle joint torque histories had minimal effect on the arms' movement (Figures 2, 3, 5). We concluded that the main effect of the increased ankle torque during the 0G attached model simulation was the control of the body's state (Figures 2, 4, 6).

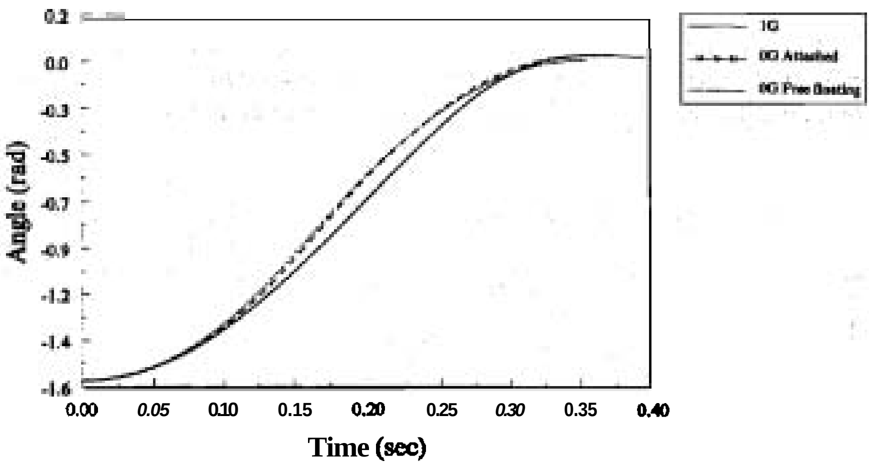

Figure 3. Angular displacement of the arms segment. The 1G condition reached horizontal at the same time as the other conditions, but total task time was increased relative to the 0G conditions.

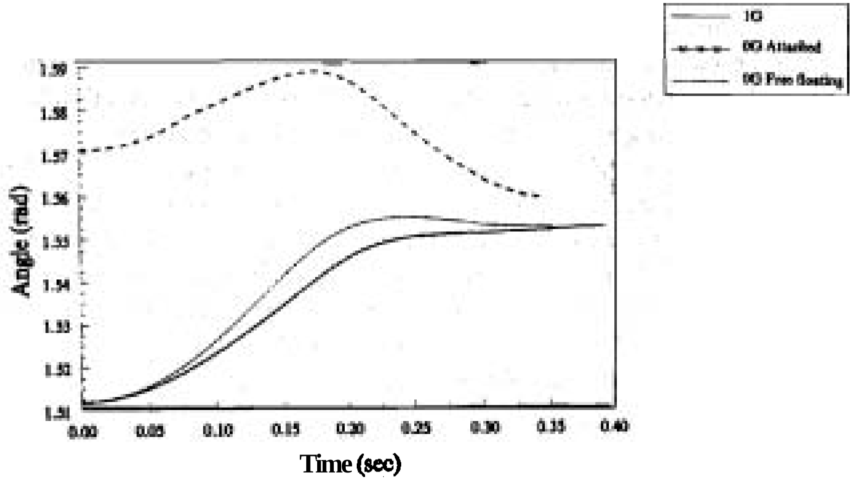

Figure 4. Angular displacement of the body segment. Although the initial position of the 0G free floating model is of no importance, notice that its final position was slightly forward, unlike the other conditions

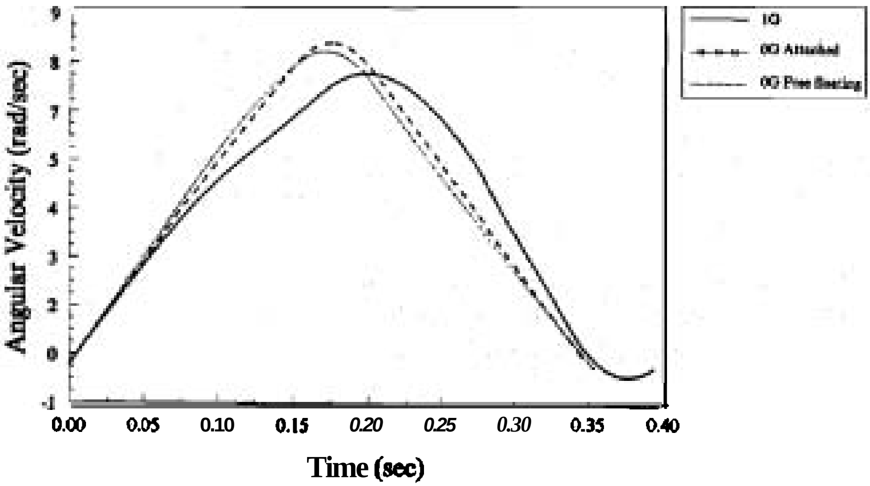

Figure 5. Angular velocity of the arms. The 0G conditions produced the highest arms velocity. Notice that the 1G condition was also able to satisfy the arms' velocity final-state requirement nearly at the same time as the other conditions.

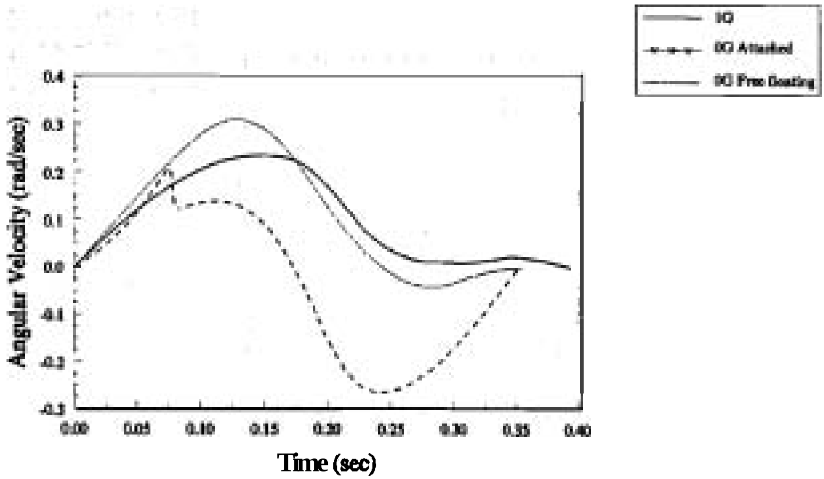

Figure 6. Angular velocity of the body. The body segment experienced low angular velocities for all conditions. Notice the sudden drop in the free floating condition at maximum plantarflexion.

During the 1G condition, the positive acceleration of the arms and the control of the body became laborious. The additional gravitational forces acting on the masses combined with the possibility of the feet coming off the ground, particularly during dorsiflexion, further constrain the model's performance. Although, during the 1G solution the feet did not come off the ground (Figure 7) and the arms segment reached the horizontal at a similar time with the other conditions (Figure 3), the body segment took longer to satisfy the necessary final states (Figures 4, 6). The differences observed during the later stages of movement in the joint torque histories between the 1G and 0G attached condition are attributed to the ineffectiveness of the ankle torque to control the massive body under gravity. It should be noted that during the braking of the arm, due to the nature of the task, the gravitational force assists the shoulder joint torque.

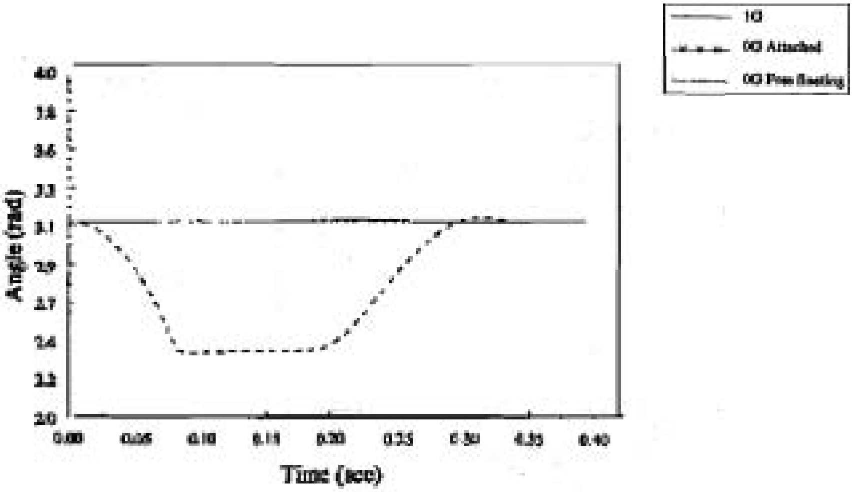

Figure 7. Angular displacement of the feet. There was no movement during the 0G attached model simulation and almost none during the 1G model simulation. The free floating simulation used maximum plantar flexion early in the task (0.075 secs).

In order to closely match human subject quiet stance in the presence of gravity and ground, and to allow for larger dorsiflexion torques, we chose the initial position of the body to be slightly leaning forward, so that the center of pressure (COP) was under the middle of the foot (Figure 8). Previous examination of two segment optimal solutions with the initial COP position under the ankle joint (body vertical) revealed similar segment kinematics, but reversed ankle joint torque patterns (Kalakanis and Abraham, 1996). This indicates that small changes in the initial stance of our human subjects could account for some of the experimental variability we observe in the electromyographic activity of the shank muscles.

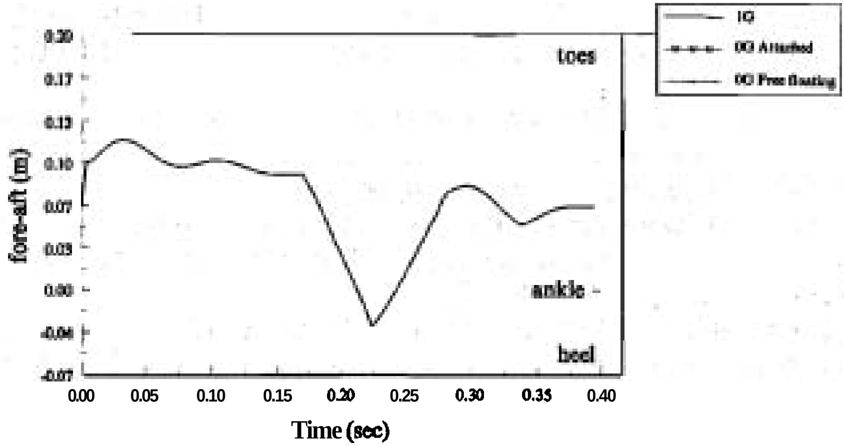

Figure 8. COP fore-aft displacement for the 1G solution. Consistent with' human subject experiments the COP for the 1G condition stayed well within the boundary of the foot during the arms movement.

CONCLUSION

Our model was simple. It did not include excitation/activation dynamics, muscle dynamics (force-velocity-length relations), geometry of joints, line of action of the muscles, knees, hips, and other parameters which could have improved its fidelity. However, it allowed us to effectively study the influence of external forces on its optimal coordination. The simulations performed indicate that manipulating the gravitational environment and the availability of ground reaction forces require adaptations in segmental coordination to optimally complete the task. It is plausible to suggest that when humans encounter the microgravity environment of space flight they also must learn new coordination modes which enable them to improve their performance. Our results show that mathematical modeling of human movement can lead to important insights into how ground reaction forces and changing levels of gravity influence movement coordination.

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