HUMANS MAKE NEAR-OPTIMAL ADJUSTMENTS OF CONTROL TO INITIAL BODY CONFIGURATION IN VERTICAL SQUAT JUMPING

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Abstract—We investigated adjustments of control to initial posture in squat jumping. Eleven male subjects jumped from three initial postures: preferred initial posture (PP), a posture in which the trunk was rotated 18° more backward (BP) and a posture in which it was rotated 15° more forward (FP) than in PP. Kinematics, ground reaction forces and electromyograms (EMG) were collected. EMG was rectified and smoothed to obtain smoothed rectified EMG (*srEMG*). Subjects showed adjustments in *srEMG* histories, most conspicuously a shift in *srEMG*-onset of rectus femoris (REC): from early in BP to late in FP. Jumps from the subjects' initial postures were simulated with a musculoskeletal model comprising four segments and six Hill-type muscles, which had muscle stimulation (*STIM*) over time as input. *STIM* of each muscle changed from initial to maximal at *STIM*-onset, and *STIM*-onsets were optimized using jump height as criterion. Optimal simulated jumps from BP, PP and FP were similar to jumps of the subjects. Optimal solutions primarily differed in *STIM*-onset of REC: from early in BP to late in FP. Because the subjects' adjustments in *srEMG*-onsets were similar to adjustments of the model's optimal *STIM*-onsets, it was concluded that the former were near-optimal. With the model we also showed that near-maximum jumps from BP, PP and FP could be achieved when *STIM*-onset of REC depended on initial hip joint angle and *STIM*-onsets of the other muscles were posture-independent. A control theory that relies on a mapping from initial posture to *STIM*-onsets seems a parsimonious alternative to theories relying on internal optimal control models. © 2013 IBRO. Published by Elsevier Ltd. All rights reserved.

Key words: EMG, muscle coordination, musculoskeletal model, optimal control, control scheme.

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Abbreviations: BP, backward rotated initial posture; CE, contractile element; CM, center of mass; EMG, electromyogram; FP, forward rotated initial posture; GAS, gastrocnemius; GLU, gluteus maximus; HAM, hamstrings; HAT, head, arms and trunk; MTCs, muscle–tendon-complexes; PEE, parallel elastic element; PP, preferred initial posture; REC, rectus femoris; RMS, Root Mean Square; SD, standard deviation; SEE, series elastic element; SOL, soleus; srEMG, smoothed rectified EMG; VAS, monoarticular vasti.

INTRODUCTION

When humans are instructed to perform a maximum height squat jump they do so from an individually preferred initial posture. In many sports situations, however, time constraints force humans to jump from other initial postures as well. This warrants the question of whether and how humans adjust their control to initial posture. In a previous study (Bobbert et al., 2008), we investigated the ability of humans to perform squat jumps from initial postures in which the center of mass (CM) was up to 20 cm higher or 20 cm lower than in the preferred posture. It was found that even without practice the subjects were able to perform successful jumps. They adjusted their control to initial squat depth, primarily by shifting the activation onset of the plantar flexors relative to the activation onsets of proximal muscles; the lower the initial height of CM, the later the activation onset of the plantar flexors occurred. Simulations with a model comprising four body segments and six Hill-type muscles showed that shifts in onset of the plantar flexors contributed substantially to maximizing jump height from the different initial postures. These findings led us to conclude that the adjustments observed in human subjects did in fact contribute to jumping as high as possible from each initial posture. It was speculated that a simple mapping from initial body posture to activation onsets of muscles is sufficient to perform near-maximum jumps (Bobbert et al., 2008; Bobbert, 2010).

Athletic subjects may actually have performed jumps from initial postures varying in squat depth in sports situations. Are they also able to make effective adjustments of control to postures to which they are less familiar? In the present study we had subjects perform squat jumps from initial postures that substantially differed in initial configuration rather than in initial height of CM, with the purpose of gaining further insight into how humans adjust control to initial posture. To assess how effective the subjects were in adjusting control, we simulated their squat jumps with a musculoskeletal model that had muscle stimulation (*STIM*) as only independent input. In general, these types of simulations help us understand the problems that the central nervous system is faced with, and allow us to explore possible control strategies. Specifically, the simulations in this study were used to answer three questions: (1) How important is it to adjust control to initial posture? (2) If the stimulation onset of only one muscle is allowed to vary with initial posture, which

muscle is the best choice? (3) How effective is a simple mapping from initial configuration to *STIM*-onsets?

EXPERIMENTAL PROCEDURES

Outline of experimental procedures

Eleven male subjects participated in this study, all of whom practiced more than twice a week sports that involved vertical jumping. The experiments were conducted in accordance with the Declaration of Helsinki, the study was approved by the local ethics committee, and all procedures were carried out with the adequate understanding and written consent of the subjects. Characteristics of the group of subjects (mean ± standard deviation) were as follows: age 24 ± 3 yr, height 1.83 ± 0.10 m, body mass 77.3 ± 7.6 kg.

The experimental session for each subject started with preparatory instructions and measurements. First, the subject was instructed to assume a squatted posture with heels off the ground and hands on the hips, and to jump as high as he could without making a countermovement. After practicing jumping under these instructions, the subject was asked to assume his preferred initial posture and perform a maximum height squat jump, five times in a row. From kinematic data collected during these five jumps, the average preferred initial posture, henceforth referred to as PP, was determined. Starting from PP, two other postures were defined. The idea was to have the same height of CM in these postures as in PP, but to have the upper body rotated either 15° more backward (BP) or more forward (FP) than in PP. The adjustments of the orientations of the other body segments required to keep CM at the same initial height were calculated off-line; in these calculations we kept the foot angle unchanged. This way, we ended up with three postures for the subject: BP, PP and FP.

After these preparations the actual experiment started, in which the subject performed three jumps from each of the three different initial postures, imposed in random order, with consecutive jumps separated by two minutes of rest. In order to make the subject perform a jump from a selected posture, we implemented this posture in a four-link template with subject-specific dimensions and held it next to the subject. Guided by feedback from the experimenter, the subject matched the posture of the template and then jumped as high as he could. Ground reaction forces were measured with a force-platform, sagittal-plane positional data of anatomical landmarks were monitored, and electromyograms (EMG) were recorded from six muscles of the right lower extremity. Jump height, defined as the difference between the height of CM at the apex of the jump and the height of CM when the subject was standing upright with heels on the ground, was calculated from the positional data. Details on the setting of the initial postures and on data collection and processing are provided below.

Collection and processing of data

Ground reaction forces were measured using a force platform (Kistler 9281B, Kistler Instruments Corp., Amherst, NY). The output signals of the platform were amplified (Kistler 9865E charge amplifier, Kistler Instruments Corp., Amherst, NY), sampled at 200 Hz, and processed to determine the fore-aft and vertical components of the reaction force and the location of the center of pressure.

For kinematic analysis, infrared light emitting diodes were placed on the right side of the body at the acromion, greater trochanter, lateral epicondyle of the femur, lateral malleolus, and fifth metatarsophalangeal joint. Together, these markers defined the positions of four body segments: head, arms and trunk (HAT), thighs, shanks and feet. During jumping, the markers were monitored in 3D using two OPTOTRAK 3020

Position Sensors (Northern Digital, Waterloo, Ontario) operating at 200 Hz. Only sagittal plane projections were used in this study. The time histories of marker positions were smoothed using a zero-lag 4th-order low-pass Butterworth filter with a cut-off frequency of 8 Hz. The locations of the mass centers of thighs, shanks and feet were estimated by combining the landmark coordinates with results of cadaver measurements presented in the literature (Clauser et al., 1969). As described elsewhere (Bobbert et al., 1996) we determined the location of the mass center of HAT relative to the two markers defining this segment from kinematic and kinetic data obtained from two equilibrium postures, one upright and one in which the hips were flexed and the upper body was oriented horizontally. With this information, the location of CM was calculated in all other body configurations found during jumping.

To record EMG from the muscles of the right leg, pairs of Ag/AgCl surface electrodes (Medicotest, blue sensor, type: N-00-S) were applied to the skin overlying gluteus maximus (GLU), long head of biceps femoris as one of the hamstrings (HAM), rectus femoris (REC), vastus lateralis as one of the monoarticular vasti (VAS), caput mediale of gastrocnemius (GAS) and soleus (SOL). The EMG-signals were amplified and sampled at 1000 Hz (Porti-17t, Twente Medical Systems). Off-line, they were high-pass filtered at 7 Hz to remove any possible movement artefacts, full-wave rectified, and smoothed using a bidirectional digital low-pass Butterworth filter with a 7-Hz cut-off frequency, to yield smoothed rectified EMG (*srEMG*).

To parameterize *srEMG*-signals, we first normalized the *srEMG*-signal of each muscle of each subject for the highest *srEMG*-value found in the three PP trials. Next, we detected an *srEMG*-onset for each muscle in each trial. This was a challenging task because the subjects were balancing on their toes in postures that were unfamiliar in the case of BP and FP, and as a result the EMG baseline showed considerable variation in some trials. To prevent this variation from leading to incorrect detection of onsets we automatically detected two points on the ascending slope of the *srEMG*-time history, one point at 20% of normalized *srEMG* and the other at 40% of normalized *srEMG*. We then fitted a line to these points and extrapolated this line backward in time to where the *srEMG*-level equaled the baseline, i.e. the average value observed while the subject was holding the initial posture (Bobbert et al., 2008). This method worked properly in the sense that *srEMG*-onsets detected objectively by this method corresponded well with *srEMG*-onsets detected by visual inspection. An occasional *srEMG*-signal showed multiple humps, causing the detection of *srEMG*-onset to be arguable regardless of whether it was done objectively or by visual inspection. In these cases we simply included the objectively detected *srEMG*-onset, accepting any potentially introduced variance. The method was also applied to the time history of the vertical ground reaction force (F_z) to detect F_z-onset, which we used to determine push-off duration (i.e. the time interval between F_z-onset and takeoff).

For comparisons of *srEMG*-onset patterns among BP, PP and FP, we needed a meaningful reference point in time. A kinematic or kinetic reference point, such as the instant of takeoff, is not useful because all kinematic and kinetic events occur as a consequence of changes in muscle activation that have taken place earlier in time. We therefore decided to take *srEMG*-onset of one muscle, muscle *R*, as reference. To test how the variation in *srEMG*-onsets across conditions depended on the choice of *R*, we quantified this variation by calculating the following statistic:

$$\sigma_R = \sqrt{\left(\frac{1}{((n_m - 1) \cdot n_s \cdot n_c \cdot n_r)} \cdot \sum_{m=1, m \neq R}^{n_m} \sum_{s=1}^{n_s} \sum_{c=1}^{n_c} \sum_{r=1}^{n_r} (t_{mscr} - t_{Rscr})^2 \right)}$$

(1)

where t_{mscr} was the *srEMG*-onset time of muscle *m* in subject *s* in condition *c* in repetition *r* ('repetition *r*' was used here instead of 'trial *t*' to avoid confusion with time); t_{Rscr} was the *srEMG*-onset