Neuromuscular control in landing from supra-maximal dropping height

A. Galindo,1 J. Barthèlemy,2 M. Ishikawa,1 P. Chavel,2 V. Martin,3 J. Avela,1 P. V. Komi,1 and C. Nicol2

1Department of Biology of Physical Activity, Neuromuscular Research Center, University of Jyväskylä, Jyväskylä, Finland; 2Institute of Movement Sciences, Faculty of Sport Sciences, University of the Mediterranean, Marseilles, France; and 3UBIAE-INSEM U902, University of Evry Val d’Essonne, Evry, France

Submitted 17 June 2008; accepted in final form 1 December 2008

Galindo A, Barthèlemy J, Ishikawa M, Chavet P, Martin V, Avela J, Komi PV, Nicol C. Neuromuscular control in landing from supra-maximal dropping height. J Appl Physiol 106: 539–547, 2009. First published December 4, 2008; doi:10.1152/japplphysiol.90776.2008.—The present study utilized high-impact supra-maximal landings to examine the influence of the pre-impact force level on the post-impact electromyographic (EMG) activity and, in particular, on the short latency EMG reflex (SLR) component. Unilateral-leg landings were performed in a sitting position on a sled apparatus after release from high, but individually constant dropping height. A lower limb guiding device fixed to the front of the sled seat allowed the subjects to sustain a given pre-set force level up to impact. This force level was either freely chosen or set at 20, 35, and 50% of maximal isometric plantarflexion force. EMG activity was recorded from eight major lower limb muscles. It was expected that the increase in the pre-impact force level would require the intervention of a protective neural strategy during the post-impact phase that would attenuate the SLR amplitude. The ultrasonography recordings confirmed that the soleus fascicles were stretched to induce SLR. The main finding was the similarity across all tested conditions of the impact peak force and post-impact EMG activity, including the SLR response. Both observations are mostly attributed to the similar EMG levels and close force levels reached toward impact. The instruction to maintain a given pre-set force level was indeed overruled when getting close to impact. It is suggested that, in the present supra-maximal landing condition, a protective central neural strategy did occur that took into account the pre-set force level to secure similar impact loads.

Impact; pre-programmed activation; short latency stretch reflex; protective strategy; ultrasonography

Poor landing techniques are identified as a major cause of serious injuries to the muscle-tendon complex as well as to the skeletal and joint structures (4, 13, 44). To cope with high-impact landings, athletes and paratroopers are properly trained, and their multijoint pre- and postlanding strategies are specifically developed to allow effective attenuation of ground impact loads (30, 48). As recently reviewed by Santello (37), a safe and smooth landing requires from the motor system a controlled muscle activation to prepare for impact. The pre-programmed origin of preactivation is generally accepted for the landing tasks (20). Clear evidence has been produced, in both animals and humans, that preactivation is modulated as a function of the drop height (1, 11, 36, 38, 39), descent velocity (48), compliance of the landing surface (21, 29), and landing technique (31, 48). In self-initiated falls, viewing the drop height before initiating the fall seems to allow accurate estimation of the impact time (25, 26, 39) and thus of the timing of the mechanical stretch (28).

The post-impact phase is characterized by a short-latency EMG burst, but its exact origin and modulation are still matters of debate and uncertainty. Both the pre-programmed (11, 12, 20) and the reflex (8–10, 21, 38) hypotheses have received support. More recently, however, the use of false force platform (28) revealed robust evidence for short latency stretch reflexes (SLR) in the gastrocnemius medialis and soleus muscles as well as for a potential reflex gain modulation by signals in descending pathways triggered by the false platform. Leukel et al. (24) have recently provided further evidence that SLR amplitude can be modulated by supraspinal centers in landing tasks. They compared landings and drop jumps and observed in the former condition reduced H-reflex sensitivity and lower background EMG activities at the time of the SLR response. The authors suggested that the nervous system would adopt a preventive neural strategy. According to Dyhre-Poulsen et al. (12), the neural strategy would be adjusted as a function of the required muscle-tendon stiffness at touch down. In addition, these modulations would be initiated before touch down, suggesting the idea of a continuum from pre- to post-impact phases (24, 37).

How post-impact responses are modulated before impact remains unclear. Preactivation level could be an important factor, since voluntary activity can modulate reflex responses. Indeed, some studies have reported a linear relationship between SLR amplitude and background EMG activity (referred to as “automatic gain control”) during isometric contractions (14, 27, 40). Conversely a nonlinear relationship was found at high force levels (32, 33, 47). Similarly, a linear increase in the H-reflex amplitude (5, 6, 34, 46) is reported with increasing plantar flexion torque production up to 50% of the maximal voluntary contraction but not above this force level (43). However, these experiments were conducted in static situations. Whether such an effect of voluntary activity on SLR response occurs during landing tasks, where fine tuning of muscle-tendon stiffness is required, remains to be investigated.

Therefore, the purpose of this study was to investigate the influence of the pre-impact force level on the post-impact SLR response in a vigorous landing task. Subjects were submitted to stressful landing conditions, while maintaining isometric force levels against a “lower limb guiding device” up to impact. It was expected that the increase in the pre-impact force level, and consequently in the muscle-tendon stiffness, might require the intervention of a protective neural strategy during the post-impact phase that would attenuate the SLR amplitude.

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MATERIALS AND METHODS

Main Protocol

Subjects. Ten physically active men (means ± SD: height, 1.78 ± 0.06 m; body mass, 71.6 ± 8.8 kg; and age, 24 ± 6 yr) took part in this main protocol. The experimental setup was approved by the Ethics Committee (n° 04008-CCPRRB Aix-Marseille II), and the procedures, purpose, and risks associated with the study were explained to all the subjects before they signed a written consent form to participate. The experiments were run in accordance with the Declaration of Helsinki.

Experimental protocol. Unilateral leg landings were performed in a well controlled sitting position on a sledge apparatus (22). The sledge inclination was 26° from the horizontal position. Before the landing tests, the subjects warmed up on the sledge by performing with both legs two series of 10 consecutive rebounds to a submaximal rebound height. Rebound movement is the standard stretch-shortening cycle exercise in which the subjects, while sitting on the sledge chair, make a push-off from the sledge force plate as soon as possible after the initial contact on the plate. After the warm-up, the maximal unilateral dropping height (H100) was measured for each subject by asking them to rebound maximally, but with the right lower limb only, when dropped from progressively increasing dropping heights. It is well known that when the dropping height is increased the subsequent rebound height increases up to a certain point, after which the additional increase in dropping height results in decrease of rebound performance (2, 23). This means that the rebound performance was systematically decreased when dropped above H100. To take into account the influence of the lower limb length on the actual dropping distance under the feet, the height (H0) corresponding to the full lower limb extension was measured while the forefoot was still on the force plate (Fig. 1A). As shown on this figure, the upper wheel of the sledge seat was used as a landmark for all the measures. The net maximal dropping distance under the feet (H100 – H0) was then calculated by subtracting H0 from H100. The supramaximal dropping height used for the actual landing test (H200; Fig. 1B) was then calculated individually by using the following formula H200 = H0 + 2(H100 – H0), in which H200 corresponds to the supramaximal dropping height used for the landing tests, H0 the individual height while extending the lower limb, and H100 – H0 the maximal dropping distance under the feet above which the rebound performance was decreasing.

For the landing tests, a lower limb guiding device (custom made) equipped with strain gauge sensor was fixed to the front part of the sledge seat. This device included a forefoot support and a visual feedback that allowed the subjects to sustain a given preset plantar flexion force level during the landing fall up to impact ("extended

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Fig. 1. Schematic presentation of the landing action on the sledge apparatus. A: measurement of the H0 parameter, which corresponds to the lower limb length when fully extended. The position of the subject before release from the H200 dropping height (B), at impact with the guiding device in extended position (C), and when the guiding device was shortened against its mechanical safety stop (D). The preset force level applied to the guiding device was controlled online from the monitor. E: identification of the measured parameters from the normal component (Fn) of the impact force-time curve.
position” in Fig. 1, B and C). After impact, the length of this device could decrease by a maximum of 11 cm up to a mechanical safety stop (“shortened position” in Fig. 1D). To set the force level individually, the subjects performed three maximal voluntary isometric plantar flexions (MVC) on the sled against the forefoot support. The test position corresponded exactly to that of the prelanding position. The highest peak force value was then taken as the individual MVC. This value was used to calculate the 20, 35, and 50% preset force levels named as P20, P35, and P50 preset conditions, respectively.

The actual landing exercise from the H200 dropping height started with a free condition followed by three preset force conditions. The free landing condition included only one trial performed with open eyes but with no visual feedback from the preset force level. The instruction was only “resist the impact.” Each of the preset force condition included one series of five trials. The total number of trials was therefore 16 per subject. The order of the three preset force conditions was randomized. The subjects received two instructions: first, to use the visual feedback to keep the preset force level constant up to impact; and second, to resist the imposed lower limb flexion.

Measured parameters. As shown in Fig. 2, the kinetic data included two impact forces: one from the sled force plate (custom-made, University of Jyväskylä, 585 × 430 × 55 mm) set perpendicularly to the lower end of the sled rails, and the second one from the guiding device to record lower plantar flexion force during the pre-impact phase until impact on the sled force plate.

EMG activity was recorded from the soleus (SOL), gastrocnemius medialis (GaM), tibialis anterior (TA), vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF), biceps femoris (BF), and semitendinosus and semimembranosus (ST-SM) muscles of the right leg using bipolar surface electrodes (Ag/AgCl, interelectrode distance of 20 mm, Biopac). Skin preparation and electrode placement followed SENIAM recommendations (18).

The sledge displacement was measured by a potentiometer (custom-made, Leine & Linde) fixed to the sled seat. Knee and ankle angular joint positions of the exercised leg were measured by means of two mono-axis electrogoniometers (Penny & Giles, Megaelectronics).

Signal processing. All signals were recorded at a 1-kHz sampling frequency (BIOPAC MP 150, 6- to 1,000-Hz bandwidth). The trigger was set to record the EMG signals with a time window that started at 800 ms before and ended at 500 ms after impact. The beginning of impact was considered as the reference (Fig. 2). The EMG signals were full-wave rectified and then low-pass filtered at 75 Hz (Butterworth type fourth-order low-pass digital filter). The obtained group averaged EMG data are shown in Figs. 3 and 4. The onset and peak latencies of the short-latency EMG burst were determined for the recorded muscles by visual inspection using a cursor on the display. The onset latency was defined as the first major deflection in the EMG record after impact (16). The subsequent EMG analysis of the mean individual data (over five trials in each preset force condition) concentrated on three different time periods: the preimpact phase, the early post-impact phase (from impact to the onset of the short-latency EMG burst), and the EMG burst period (from onset to the end of this short-latency EMG burst).

To follow the possible of EMG changes during the pre-impact phase, the EMG signals were averaged for the five trials of each subject and then group averaged by successive time periods of 100 ms with an overlap of 50 ms, in parallel with the guiding force device. Two time periods of 100 ms were compared for the SOL, GaM, VL, and VM muscles: the −600 to −500 ms that preceded for all subjects the beginning of the fall and the −100 to 0 ms that preceded the impact. Since TA, RF, and BF muscles presented only a fairly low muscle activity observed in all conditions for RF, BF, and ST-SM muscles in this early pre-impact phase, the time window 80 – 0 ms before impact was chosen for its stability and used instead as the reference (100%) for normalization of SOL, GaM, TA, VL, and VM EMG data for each subject. Because of the very low muscle activity observed in all conditions for RF, BF, and ST-SM muscles in this early pre-impact phase, the time window 80–0 ms before impact was used instead for their normalization (10).

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As reflected by the curves of Fig. 3C, this method would not have been appropriate for the SOL and GaM muscles since both of them were...
characterized at that time by a clear reduction in EMG activity. Co-activation between TA and SOL in the last 100 ms before impact was calculated as a ratio of the mean antagonist (TA) activity to the mean agonist (SOL) activity.

For the so-called EMG burst period, SOL and GaM timing (onset and peak of the short-latency EMG burst), and mean amplitude (from onset to the end of the short-latency EMG burst) were analyzed for each trial and then averaged per tested condition. The time window used for TA average was 0–50 ms (10, 28). For both quadriceps (VL, VM, and RF) and hamstring (BF, ST-SM) muscle groups, the EMG burst activity was averaged from 20 ms to the end of the short-latency EMG burst.

To allow comparison with the falling heights used in the literature on feet-first landings, calculation of the exact dropping distance (from the lower part of the guiding device to the force plate; see Fig. 1B) was performed.

Quantification of the normal ground reaction force signal was restricted to the early post-impact phase. This included the impact peak (F1) and the mean rate of loading (from the first value above the baseline to F1 (17)) as well as the minimum value reached within the early (35 ms) post-impact phase and the peak force reduction (PFR = F1 – minimum force value) (Fig. 1E) (3). The calculation of peak force reduction was used to quantify the impact absorption. For example, a large peak force reduction should be seen as a large

Fig. 3. A: group averaged values of the planatarflexion force level produced up to impact in the three preset force conditions (P20, P35, and P50, expressed in percentage of the maximal voluntary contraction) and in the freely chosen (FREE) force condition. B: group averaged values of the ankle angle in all conditions during the pre- and post-impact phases. C: Corresponding group averaged SOL, GaM, and TA EMG activities when expressed in percentage of their mean activity before release (600 to 500 ms) in the P20 condition. For all traces, the impact is represented by a bold and broken line.

Fig. 4. A: group averaged values of the knee angle during the pre- and post-impact phases in the three preset force conditions (P20, P35, and P50, expressed in percentage of the maximal voluntary contraction) and in the freely chosen (FREE) one. B: corresponding group averaged VL and VM EMG activities when expressed in percentage of their mean activity before release (~600 to ~500 ms) in the P20 condition. Because of the very low activity of the BF muscle in this early pre-impact phase, the time window 80–0 ms before impact was used for its normalization. For all traces, the impact is represented by a bold and broken line.
impact absorption. The impact peak was normalized relative to the subject body weight, and the peak force reduction was normalized relative to F1.

Knee and ankle angular positions were averaged along the pre-impact phase by successive 100-ms time windows with an overlap of 50 ms until impact. Analysis of the post-impact phase was restricted to the first 50 ms. Impact-induced knee and ankle angular variations were thus calculated by the values from impact to 50 ms. Peak sled velocity during the last 10 ms before impact was measured by the differentiation of the sledge displacement signal.

Additional Protocol

Nine physically active men (means ± SD: height, 1.80 ± 0.05 m; body mass, 78 ± 4 kg; and age, 22 ± 3 yr) took part in this additional protocol. To examine further the hypothesis of a stretch-reflex occurrence in the triceps surae muscle group, ultrasonography recordings of the SOL muscle were added to the main protocol. In this additional protocol, the EMG recordings were restricted to the SOL and GaM muscles and subsequently analyzed from −200 ms before impact to 60 ms post-impact. Otherwise, the experimental setup was the same as in the main protocol. All force data (from the force platform and the guiding device) and joint angle parameters were recorded similarly as in the main protocol.

Ultrasonography recording of the SOL fascicle length and fascicle pennation angle during each trial during the last 200 ms of the landing fall and during the entire post-impact phase was added to evaluate how the muscular tissue responded to high-impact loads. The ultrasound unit (SSD-5500SV, Aloka) was set to operate at 96 Hz. The probe (Linear Probe 5.0/10 MHz wave frequency with 60-mm scanning length) was fixed below the proximal myotendinous junction of GaM. In the analysis, the fascicle length was tracked frame by frame between the superficial and the deep aponeurosis of SOL. The pennation angle was determined between the superficial aponeuroses and the line drawn to the fascicle. The ultrasonography data were reported for the entire 200 ms pre-impact, for the early post-impact (0–35 ms), and for the post-impact (35–50 ms) periods. For the SOL muscle, the turning point from fascicle shortening to stretching during the early post-impact phase was taken as the indication of a mechanical stretch susceptible to result in SLR response. The mean stretching rate was calculated between the minimum and the maximum values of SOL fascicle length in the 0- to 50-ms period of the post-impact phase.

Statistics

The results are presented as means ± SD. A one-way ANOVA with repeated measures (Statistica 6, Statsoft) was used to test the main effects of the preset force condition on each measured variable from the main and the additional protocol. The level of statistical significance was set at \( P < 0.05 \). In case of significant effect, Newman-Keuls post hoc test was used to test the interactions.

RESULTS

Main Protocol

The distance measured between the guiding device support and the force plate before release from the \( H_{300} \) dropping height corresponded to 63.6 ± 11.8 cm. The fall phase lasted 470 ± 26 ms, reaching a peak velocity of 2.5 ± 0.3 m/s.

Pre-impact phase. Analysis of the pre-impact phase over a 600-ms time period was used 1) to check how the three target preset force levels (P20, P35, and P50) were followed and 2) to compare them to the freely chosen one.

Before release (from −600 to −500 ms), the preset force levels were well maintained and differed among the three tested conditions (P20, 19 ± 1%; P35, 34 ± 1%; P50, 48 ± 1%; \( P < 0.001 \)). The freely chosen force level (12 ± 6%) was on the average lower than all the other ones (\( P < 0.001 \)) (Fig. 3A). The respective ankle (Fig. 3B) and knee (Fig. 4A) joint angle analysis did not reveal any significant influence of the preset force level. The ankle joint was only in the FREE condition less plantar-flexed than in the P35 condition (\( P < 0.05 \)) (Fig. 3B). The EMG analysis demonstrated that the higher the preset force level, the higher the overall lower limb muscle activity was in SOL, GaM, and TA (Fig. 3C), and in VM and VL (Fig. 4B). The FREE testing condition was characterized for each of these muscles by a lower activity than in the P50 condition (\( P < 0.01 \)). As shown in Fig. 4B for the BF muscle, the activities of the hamstring muscle group (BF and ST-SM) and of the RF muscle were very low up to impact.

After release, the subjects did not succeed in keeping the preset force level constant as instructed. Both P20 and P35 conditions were characterized by a progressive increase in plantar flexion force of 57 and 17%, respectively (Fig. 3A). The largest increase was observed in the FREE condition (112%). At impact, however, the recorded force levels were still similarly ranked and significantly different among all conditions (\( P < 0.001 \)). At that time, no significant difference was observed in either knee or ankle joint position (Figs. 3B and 4A). The EMG analysis revealed an overall increase in muscle activity leading to similar EMG levels (no significant difference) among the preset force conditions. In terms of timing, the additional increase in SOL, GaM, VL, and VM activity became significant at 300 ms before impact and for TA at 250 ms (\( P < 0.05 \)). SOL and GaM EMG activities peaked in the 150- to 50-ms time period before decreasing slowly (Fig. 3C), whereas TA, VL, and VM increased up to impact (Figs. 3C and 4B). The FREE condition also showed large EMG increases and became close in magnitude to the values found in the other testing conditions. Only in the SOL muscle was the FREE condition value lower than the others during the last 100 ms before impact (Fig. 3C). Despite a large interindividual variability, TA activity was significantly higher than in the P50 condition (\( P < 0.05 \)), but the TA/SOL co-activation analysis did not reveal any difference across conditions.

Post-impact phase. As illustrated in Fig. 5A, the preset force level was not found to influence significantly the first impact peak force (F1) either in absolute or in relative values. Across the tested conditions, F1 timing was as short as 4.6 ± 0.7 ms, and its amplitude (tolerated by a single leg) averaged five times body weight. The loading rate did not differ either among conditions. However, a larger peak force reduction was found in the FREE condition (\( P < 0.01 \)) compared with each of the preset force conditions (Fig. 5B). The lowest force value was reached in all tested conditions at −10 ms.

As illustrated in Fig. 5C, no knee and ankle angular changes were detectable within the first 10 ms. In terms of mean angular velocities, the values measured at the ankle joint were significantly higher than the ones measured at the knee joint (\( P < 0.01 \)). Due to high interindividual variability, however, no significant difference was found among the tested conditions for the knee (P20, 120 ± 46°/s; P35, 120 ± 51 °/s; P50, 109 ± 47 °/s; FREE, 109 ± 35 °/s) and the ankle (P20, 188 ± 34 °/s; P35, 193 ± 31 °/s; P50, 205 ± 39 °/s; FREE, 173 ± 65 °/s) joints. The same was true when the comparison was applied to the first 10 ms post-impact.
With regard to the EMG activity, the early post-impact phase was characterized by a stable EMG activity in both SOL and GaM (0–35 ms) as well as in BF and in the quadriceps (VL, VM, and RF) muscle group (0–20 ms). Independently of the tested condition, this was followed in each of these muscles by a systematic EMG burst (Figs. 3C and 4B). The EMG burst timing was found to be particularly stable among conditions, especially for SOL, GaM, VM, and VL (Table 1). Similarly, the mean EMG burst amplitudes differed among muscles but not across conditions. The inter-individual variability was particularly high for the BF and ST-SM muscles. As shown in Fig. 3C, the EMG burst amplitude was much larger than the background (pre-impact) EMG activity in the SOL than in the GaM muscle. The same was true compared with their respective “maximal” activity recorded in the MVC test. Due to a large individual variability in the level of intervention of the quadriceps muscle group during the MVC test, the same comparison was not performed for this muscle group.

Additional Protocol

Pre-impact phase. The kinetic, kinematics, and EMG analysis revealed significant and similar patterns of changes toward impact compared with the main protocol results.

The ultrasonographic analysis of the last 200 ms before impact revealed initially longer SOL fascicle lengths in the FREE and P20 conditions compared with the other two conditions (P < 0.05). Furthermore, in all tested conditions, a progressive decrease in SOL fascicle length associated with an increasing pennation angle (Fig. 6). More specifically, the change in fascicle length during this time period was greater (P < 0.05) in the P20 (−0.44 ± 0.13 cm) and P35 (−0.47 ±

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Table 1. Latencies of the EMG burst onset and peak values

Values are means ± SD from 10 subjects. Latencies of the electromyographic (EMG) burst onset and peak values when recorded during the postimpact phase from the soleus (SOL), gastrocnemius medialis (GaM), vastus lateralis (VL), and vastus medialis (VM) muscles. T0, the exact beginning of the rise of the normal ground reaction force; P20, P35, P50, the preset force level conditions at 20, 35, and 50%, respectively, of maximal isometric plantarflexion force; FREE, the freely chosen force condition. Timing of the VL and VM EMG bursts in the FREE condition are not mentioned due to large inter-individual variability.
0.28 cm) conditions than in the P50 one (−0.20 ± 0.17 cm). At impact, SOL fascicle length was still longer in the FREE and P20 conditions compared with the other two conditions (P < 0.05). Parallel analysis of the SOL pennation angle revealed a larger increase in the P20 (+4.2 ± 1.6°) and FREE (+4.0 ± 1.6°) conditions compared with the P50 (+1.9 ± 1.0°) condition (P < 0.05). At impact, SOL pennation angle in the FREE condition was smaller than in the P50 condition (P < 0.05) (Fig. 6).

**Post-impact phase.** Similarly to the main protocol, analysis of the SOL and GaM EMG bursts confirmed their stability in both timing and mean amplitude and independently of the preset force condition.

The impact-induced ankle dorsiflexion resulted in a significant increase in SOL fascicle length (P < 0.05) in the early post-impact phase (up to 35 ms) (Fig. 6) that preceded the EMG burst by ~15 ms. In this phase, SOL fascicle length remained longer in the P20 and FREE conditions than in the P35 and P50 ones (P < 0.05). Thereafter, all conditions got closer to each other so that from 35 to 60 ms post-impact the mean fascicle length stretching rates did not differ among conditions (Fig. 6).

**DISCUSSION**

The present study utilized landings from supramaximal dropping height to study the influence of the pre-impact force level on the subsequent post-impact EMG response and, in particular, on the short latency EMG reflex (SLR) component. The main finding was the similarity across all the testing conditions of the post-impact EMG activity, including the short latency EMG burst response.

The post-impact EMG pattern observed in our experiment was very similar to the one observed in “feet-first” human vertical landing performance (10, 12, 25, 28, 37). This pattern was characterized by a sharp EMG burst starting at 35 ms after impact for the triceps surae (Fig. 3C) and at 20 ms for the quadriceps (Fig. 4B) muscle group (Table 1). Similarly to the results of Dyhre-Poulsen et al. (12), the antagonist TA activity was found to increase toward impact and to show a post-impact alternating pattern of bursts and pauses with the agonist (SOL) muscle. The hypothesis of a short latency stretch reflex (SLR) intervention is supported for the triceps surae muscle group by the present SOL and GaM EMG burst latencies (Table 1) that are close to those of the literature (42, 45) as well as by the stretch amplitude and velocity of the SOL fascicle length revealed by the ultrasonography technique (7). The observed behavior of the SOL fascicle length is quite similar to the one reported in drop jumps (41), in which the impact induced a clear turning point from shortening to lengthening (Fig. 6). On the other hand, the intervention of a protective neural strategy in all testing conditions is suggested by the stability of the SLR amplitude of both SOL and GaM muscles across the FREE and the three preset force conditions (Fig. 3C). This differed from our expectations that the automatic gain control would lead to a progressive increase in SLR amplitude from P20 to P35; a trend that was expected to be attenuated in the most stressful (P50) condition. For the quadriceps muscle group, the same stability was observed in the EMG burst, but its origin is more questionable. The electromyographs did not detect any significant knee angular flexion before the burst in VL and VM EMG recordings. Ultrasonography recordings are missing.
however, to confirm their absence at the muscle fascicle level and, thus, at the muscle spindle level.

Among the potential factors that could have influenced the SLR amplitude, reciprocal inhibition from increased TA EMG was expected to occur at high preset force levels. This hypothesis is not supported, however, by the absence of significant difference across the preset force conditions. Only the FREE testing one presented a higher TA pre-impact activity, but this did not affect significantly the TA/SOL co-activation level (Fig. 3C). This minor influence of the TA co-activation may be partly attributed to the presence of the lower limb guiding device that enhanced the stability and the security of the ankle joint position before and after impact. The increase of the preset isometric contraction level was also expected to result in a progressive suppression of Ib inhibition due to descending control (35). A gating mechanism has been reported, however, that would still allow transient inhibitory potentials in case of rapid increases in contraction force (19, 35). In the present case, this transient influence should not have differed among conditions since the impact peak did not differ among them (Fig. 5). The exact influence of Ib inhibition on the present SLR response remains, however, questionable. On the other hand, the low and similar levels of SOL background EMG observed in all conditions during the early post-impact phase (Fig. 3C) are in agreement with Leukel et al. (24) of preprogrammed influence on presynaptic Ia inhibition. In their study, this was based on the observation of lower H reflex and background EMG activity during the post-impact phase in landing than in rebound. In the present study, however, this argument does not apply to the other lower limb extensor muscles that were found to remain at higher EMG levels (close to their activity level at impact). This leads us to the major observation at impact of similar EMG levels across conditions, with closer plantar-flexion force levels than expected. The first instruction to maintain a given preset force level was indeed systematically overruled toward impact. It is suggested that, despite a constrained pre-impact phase, protective neural strategies might have attenuated the initial differences toward impact.

Furthermore, it seems that the pre-impact neural strategy was adjusted before release to the maintained force level in both FREE and preset force conditions. This is supported by the fact that the timing of the additional rise in EMG was earlier than the one reported in vertical falls during the pre-impact phase (10, 38, 39). In addition, the EMG build-up and the produced pre-impact force occurred at a faster rate when starting from a lower preset force level. Finally, both SOL and GaM muscles were characterized by a decrease in EMG activity during the last 150 ms preceding impact (Fig. 3C). Based on the landing and rebound literature (12), this particular EMG pattern would not aim to resist maximally but, instead, to damp the approaching high-impact loads. This global neural strategy led to closer force levels at impact than what was planned with this protocol. This is considered the major explanation for the absence of significant intercondition difference in the impact peak F1 (Fig. 5A) and subsequent SOL stretch velocity (Fig. 6). Intervention of a protective neural strategy is also supported by the fact that the present landing exercise was particularly stressful. The time to reach the initial peak force (F1) was extremely short, 4.6 ms compared with 53 ms in feet-first vertical landing (38), with a sharp and large impact force (~5 times body weight on one leg vs. 5.6 times body weight on two legs). The present long duration of the fall (460 vs. 390 ms in vertical landing) should have favored also the intervention of a protective neural strategy in the adjustment of the tendino-muscular stiffness toward impact.

The FREE testing condition was considered as a more natural way to prepare for the high impact loads. Supporting the damping hypothesis, the FREE condition was for any of the measured EMG and force parameters closer to the P20 than to the P50 condition. Although the impact peak force (F1) did not differ among conditions, P20 and FREE were indeed characterized by larger peak force reduction (Fig. 5B). This parameter has been suggested to reflect changes in active stiffness properties of the muscle (3) so that a larger peak force reduction should be seen as a larger impact absorption. Interestingly, since the immediate post-impact EMG activity did not differ among conditions, the larger peak force reduction observed in P20 and FREE may thus be mostly attributed to their lower pre-impact force levels. This is further confirmed by the longer SOL fascicle length measured up to impact in these conditions (Fig. 6). A lesser involvement of other plantar-flexor muscles might be expected as well in these two conditions of lower preset force levels (15).

In conclusion, the results of this specific maximally landing task support, on the one hand, a centrally controlled activation of the different muscle groups. On the other hand, it also confirms the occurrence of a SOL SLR response triggered by fascicle stretching. Differing from our expectations, the different preset force levels before impact did not influence either the timing or the amplitude of the post-impact EMG burst response. The fact that the neuromuscular system “overruled” the instructions to keep the preset force level might reflect the intervention of a protective neural strategy. In agreement with Santello (37), the natural protective neural strategy in the case of high dropping height may thus be considered as a continuum from the pre- to the post-impact phase to prevent overload of the tendino-muscular system.

ACKNOWLEDGMENTS

We are particularly grateful to Professor Gabriel Gauthier for constructive comments and technical help during this work. We also express our gratitude to Professor C. Brunet for support during the experiment.

GRANTS

The study was supported in part by grants from the Ministry of Education (no. 88/627/2005) Finland and from the HUMOS 2 project (G3RD-CT-2002-00803) of the European Commission.

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