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Leg stiffness can be maintained during reactive hopping despite modified acceleration conditions

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ABSTRACT

Aim: The aim of the present study was to evaluate reactive hops under systematically modified acceleration conditions. It was hypothesized that a high preactivity of the leg extensors and phase-specific adjustments of the leg muscle activation would compensate the alterations caused by the various acceleration levels in order to maintain a high leg stiffness, thus enabling the jumper to perform truly reactive jumps with short ground contact times despite the unaccustomed acceleration conditions.

Methods: Ground reaction forces (GRF), kinematic and electromyographic data of 20 healthy subjects were recorded during reactive hopping in a special sledge jump system for seven different acceleration levels: three acceleration levels with lower than normal gravity (0.7*g*, 0.8*g*, 0.9*g*), one with gravitational acceleration (1*g*) and three with higher acceleration (1.1*g*, 1.2*g*, 1.3*g*).

Results: The increase of the acceleration from 0.7*g* to 1.3*g* had no significant effect on the preactivity of the leg extensors, the leg stiffness and the rate of force development. However, it resulted in increased peak GRF (+15%), longer ground contact time (+10%) and increased angular excursion at the ankle and knee joints (+3°).

Discussion: Throughout a wide acceleration range, the subjects were able to maintain a high leg stiffness and perform reactive hops by keeping the preactivity constantly high and adjusting the muscle activity in the later phases. In consequence, it can be concluded that the neuromuscular system can cope with different acceleration levels, at least in the acceleration range used in this study.

Introduction

The stretch–shortening cycle (SSC) has been studied extensively (Bosco et al., 1982; Finni et al., 2001; Komi and Gollhofer, 1997). It can be defined as the stretching of a preactivated muscle–tendon unit (MTU), immediately followed by a shortening action of the muscle (Komi, 1984). The SSC is characterized by a high efficiency due to a potential storage of energy in the elastic elements of the MTU (Asmussen and Bonde-Petersen, 1974; Gollhofer, 1987; McCaulley et al., 2007) and is often described as the natural type of muscle action, as it is constitutive for everyday activities such as running or jumping, as well as for many sports. Several studies suggest that the prerequisites for a storage of energy in the MTU during reactive jumps using the SSC are the absence of heel contact (Schmidtbleicher and Gollhofer, 1982; Bobbert et al., 1987), and above all a high muscle stiffness, which in turn requires a high preactivation of the leg extensor muscles (Ishikawa and Komi, 2004; Komi and Gollhofer, 1997). If these requirements are not met, the strain energy cannot be stored in the elastic components of the MTU and instead dissipates as thermal heat. In consequence, many studies have tried to identify optimal conditions that allow the execution of reactive jumps with high leg stiffness. It has been shown that a number of factors determine the range of optimal conditions for reactive jumps: these factors include the falling height (Bosco and Komi, 1979; Kyröläinen and Komi, 1995; Schmidtbleicher and Gollhofer, 1982), the additional mass that is attached to the jumper (Bosco and Komi, 1979; Gollhofer and Kyröläinen, 1991) and the kinetic energy just prior to ground contact, which is often considered to be the most important factor, as it includes both the mass and the falling height (Bubeck, 2002).

One factor that seems to limit the possibility to perform reactive jumps to a particularly narrow range is the acceleration acting on the jumper: there are only a few studies, but they all suggest that jumps with accelerations both above 1*g* and below 1*g* lead to inferior results compared to jumps with gravitational acceleration. For example, Avela et al. (1994) employed a "lifting

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block system'' using weights and counterweights to modulate the effective acceleration acting on the jumper. They compared two different accelerations (0.8*g* and 1.2*g*) to the normal acceleration of 1*g* and observed lower ground reaction forces (GRFs), longer ground contact times (GCT) and lower preactivity of the leg extensors, both for the jumps with an acceleration of 0.8*g* and the ones with an acceleration of 1.2*g*. Consequently, the authors concluded that ''all the results emphasized considerable adaptation of the neuromuscular system to the normal gravity condition'' (Avela et al. (1994). Albeit using different systems for the variation of the acceleration, other authors came to similar conclusions (Bubeck, 2002; Cavagna et al., 1972; Gollhofer and Kyröläinen, 1991), and Gollhofer and Kyröläinen suggested that mainly the low preactivity of the leg extensors was responsible: ''the centrally programmed activity prior to the contact can be seen as the decisive mechanism in the regulation of the stiffness behavior of the tendomuscular system'' (Gollhofer and Kyröläinen, 1991). However, it remains unclear if the observed optimal jump performance at the acceleration of 1*g* is really due to an evolutionary adaptation to the gravitational acceleration, or if the systems modulating the acceleration used in the aforementioned studies were simply not adequate for the task. For example, the system used by Bubeck (2002) could vary the acceleration only by also varying the inertia of the jumper, which probably had a considerable confounding effect on the results. Likewise, the abrupt change of the acceleration at the moment of ground contact that was inherent to the systems used by Gollhofer and Kyröläinen (1991) and Avela et al., 1994 had probably also a strong effect on the results.

Therefore, the purpose of the present study was to assess the effects of modulated acceleration with a new system which was already validated in a previous study (Kramer et al., 2010) and which allows a variation of the acceleration without changing the inertia or inducing sudden changes of the acceleration acting on the jumper. It was hypothesized that a high preactivity of the leg extensors and phase-specific adjustments of the leg muscle activation would compensate the alterations caused by the various acceleration levels in order to maintain a high leg stiffness, thus enabling the jumper to perform truly reactive jumps with short ground contact times and a high rate of force development despite the unaccustomed acceleration conditions ranging from 0.7*g* to 1.3*g*.

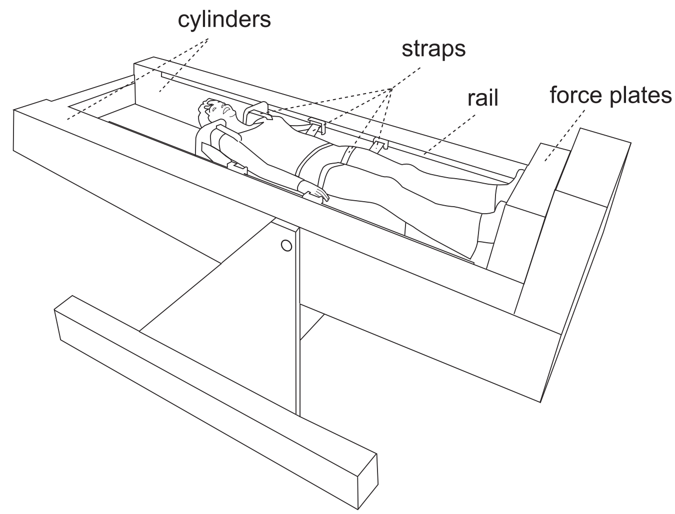

Fig. 1. Sledge jump system in the horizontal position. The participant is fixed to the wooden sledge with straps. The straps are attached to the rails and can slide in the direction of the rails. The participant stands on two force plates (separated, one for each foot). Reprinted from Kramer et al., (2010), with permission from Elsevier.

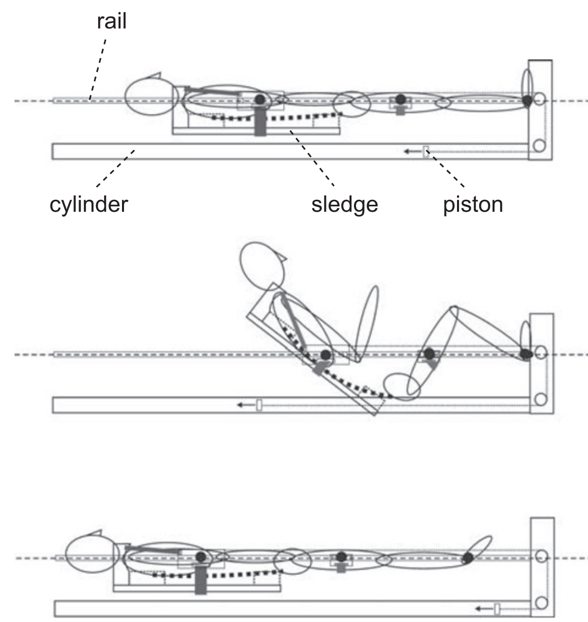

Fig. 2. Longitudinal section of the SJS. The two low-pressure cylinders pull the participant onto the force plates. The second drawing illustrates the freedom of movement in the ankle, knee and hip joint (the downward movement of the hip is not constrained by the cylinders like the drawing might suggest, as the cylinders are to the left and right of the participant, not below; see Fig. 1). Note that the figure is not intended to show a subject hopping. Reprinted from Kramer et al. (2010), with permission from Elsevier.

Methods

Subjects: Twenty subjects (4 females, 16 males) volunteered to participate in this study. The participants were healthy, physically active students at the department of sports science. All participants gave written informed consent to the experimental procedure, which was approved by the ethics committee of the University of Freiburg and in accordance with the Declaration of Helsinki. Their mean ($\pm$ standard deviation, SD) height, body mass and age were 177 ± 7 cm, 74 ± 8 kg and 23 ± 3 years, respectively.

Experimental design: A single-group repeated-measures study design was used to examine differences between hops in the SJS with 7 different accelerations (0.7, 0.8, 0.9, 1, 1.1, 1.2 and 1.3*g*). After a 10-min warm-up phase (consisting of running, tapping and hopping), the hops were performed with the instruction ''jump as stiff as possible''. One trial consisted of 40 hops with a 2-min break after 20 hops. Before each trial, the participants performed 10 hops with the acceleration of the subsequent 40 hops in order to familiarize the participants with the new acceleration condition. The hops were performed bare-footed on two force plates (Leonardo®, Novotec Medical, Germany). The ground reaction forces (GRFs) were recorded separately for the right and the left foot and synchronized to the electromyograpic (EMG) signals. The order of the test conditions was balanced between subjects to control confounding effects such as fatigue. In addition, the participants were given at least 3 min of rest in between trials.

Sledge jump system: The SJS (see Figs. 1 and 2) was developed by Novotec Medical (Pforzheim, Germany). For a detailed description, see Kramer et al. (2010). Basically, the subject can jump in the system with hardly any restrictions concerning the joint movements, allowing almost natural jumps. Since the movement direction is along a horizontal axis, the forces generated by the two low-pressure cylinders substitute the gravitational force. The pressure was adjusted in a way that the forces produced by the cylinders matched the designated fraction of the subject's weight, e.g. an acceleration of 0.7*g* for a subject with a weight of 750 N was achieved by setting the force to $750\ \text{N} \times 0.7 = 525$ N.

Kinematics: The jumps were recorded with a motion capturing system (Vicon, UK) using ten cameras (MX40, 200 Hz). The markers were placed on the following anatomical landmarks of the right leg: hallux, fifth metatarsal bone, lateral malleolus, lateral knee joint center and greater trochanter. In addition, one marker was placed on the sternum. Those markers were used to generate a 2D-model of the right leg, from which three joint angles were calculated (ankle, knee and hip). An additional marker on the sledge was used to determine the jumper's velocity.

Electromyography: Bipolar surface electrodes (Ambu Blue Sensor P, Denmark) were placed over M. soleus (SOL), M. gastrocnemius medialis (GM), M. tibialis anterior (TA), M. rectus femoris (RF), M. vastuslateralis (VL) and M. biceps femoris (BF) of the right leg. The longitudinal axes of the electrodes were in line with the presumed direction of the underlying muscle fibers. Interelectrode resistance was kept below 3 kΩ by means of shaving, light abrasion and degreasing of the skin. The EMG signals were transmitted via shielded cables to the amplifier (band-pass filter 10 Hz–1 kHz, 1000 × amplified) and recorded with 4 kHz.

Data processing: After removing DC offsets, the EMG signals were rectified. Afterwards, the means of the EMG and force signals were calculated for each trial, using the GRF of the right force plate as a trigger signal for the moment of ground contact (GC). Then, the iEMG was calculated by integrating the mean EMG signal of 4 time intervals, based on previous reported latencies and durations of the reflex components (Lee and Tatton, 1978; Sinkjaer et al., 1999): the preactivity phase (PRE) from 50 ms before ground contact (GC) until GC, the phase of the short-latency response (SLR) from 30 ms after GC until 60 ms after GC, the phase of the medium-latency response (MLR) from 60 ms until 90 ms and the phase of the long-latency response (LLR) from 90 ms until 120 ms (see Fig. 3). To ensure inter-subject comparability, the iEMG during each phase was normalized to the subject's iEMG from −150 ms to +120 ms, and the GRF was normalized to the subject's body weight. The rate of force development (RFD) was calculated as the peak force divided by the time from GC until the force signal reached its peak. The joint angles were determined at the time of GC and the angular joint excursions were calculated from GC until the GRF reached its peak. The leg stiffness was calculated according to Günther and Blickhan (2002) as the ratio of the peak GRF to the displacement of the hip marker (greater trochanter) during the time interval

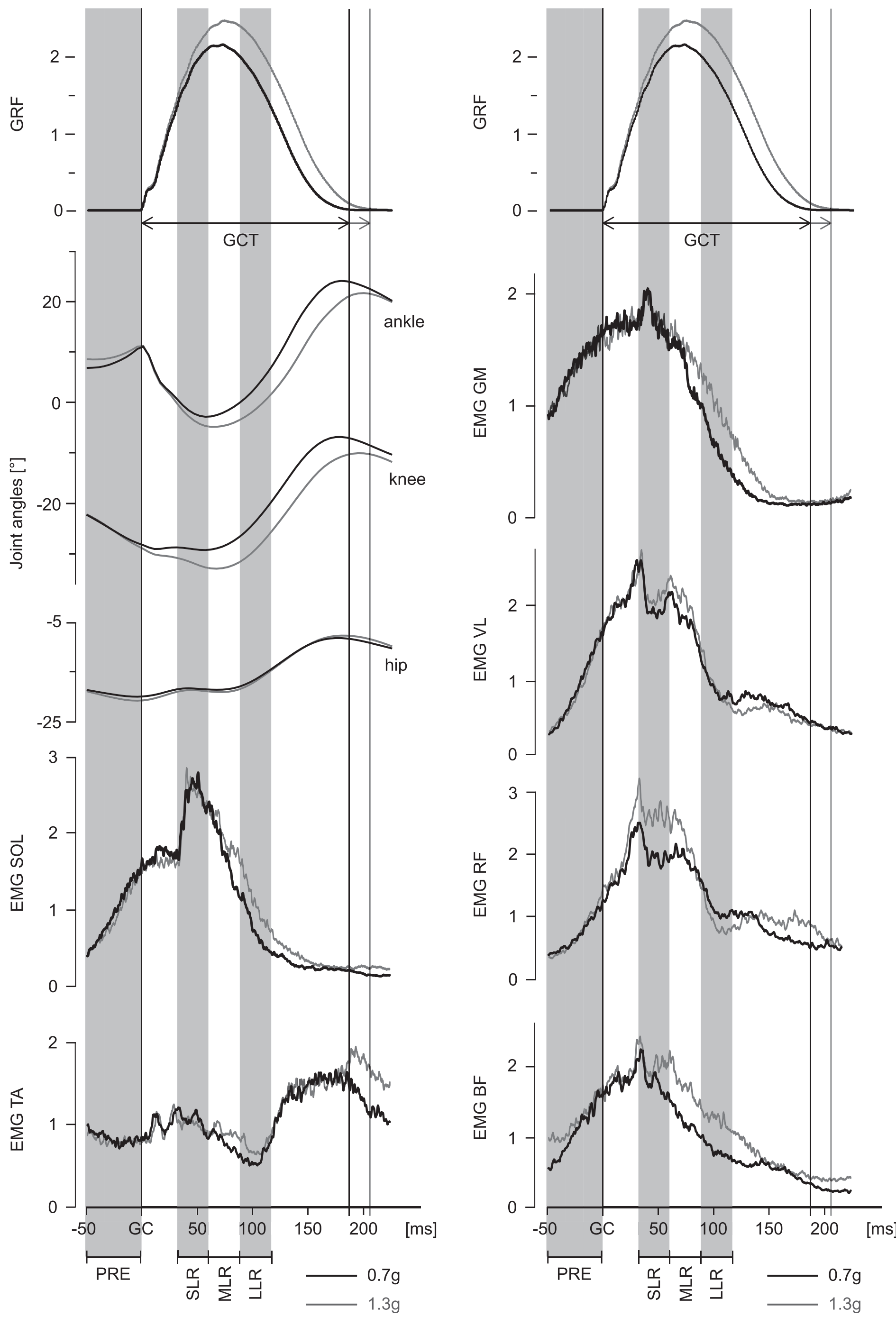

Fig. 3. Averaged ground reaction forces (GRFs, right leg), joint angles and EMG data for all participants during two acceleration levels. Black lines represent the hops with an acceleration of 0.7g, the gray ones represent the hops with 1.3g. The ground contact time (GCT) is marked as the time between ground contact and takeoff. Also marked are the relevant EMG phases: PRE 50 ms before GC until GC, SLR 30–60 ms after GC, MLR 60–90 ms and LLR 90–120 ms. For the joint angles, negative values indicate more flexion in comparison to upright stance. To ensure inter-subject comparability, the GRF was normalized to the body weight and the EMG to the average EMG activity from −150 ms to +120 ms.

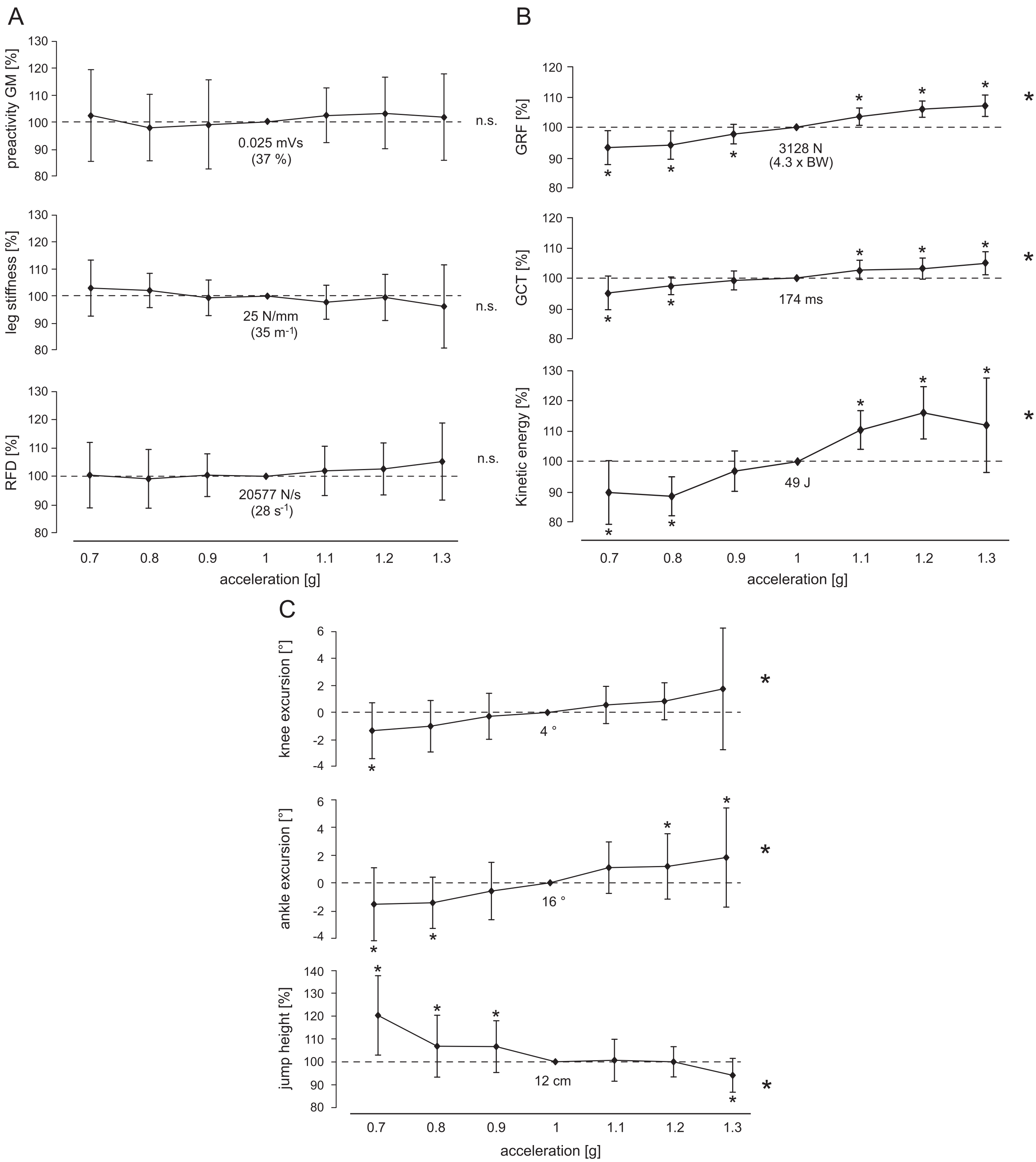

Fig. 4. A. Grand mean of the most important parameters of a reactive jump for all participants for the 7 acceleration levels from 0.7*g* to 1.3*g*: preactivity of the leg extensors (the medial gastrocnemius is shown as a representative example), leg stiffness and rate of force development (RFD), all normalized to the respective 1*g* value. For the 1*g* condition, both the absolute values and the relative values (i.e., normalized to the iEMG from −150 ms to +120 ms in the case of preactivity and to the body weight in the case of leg stiffness and RFD) are displayed. The acceleration has no statistically significant influence on any of the 3 parameters. B. Grand mean of the kinetic data for the seven acceleration levels: peak ground reaction force (GRF, both legs), ground contact time (GCT) and kinetic energy just prior to ground contact, all normalized to the respective 1g value. The acceleration has a statistically significant influence on all 3 parameters (big * symbol, denoting a significant ANOVA result). A small * symbol denotes a statistically significant difference compared to the 1*g* condition. C. Grand mean of the kinematic data for the 7 acceleration levels: knee excursion in the eccentric phase of the jump, i.e. from ground contact until the force curve reached its peak, ankle excursion in the eccentric phase of the jump and jump height. The acceleration has a statistically significant influence on all 3 parameters.

from GC until the GRF reached its peak. The kinetic energy prior to ground contact was calculated based on the sledge marker's velocity via the formula $E = 0.5mv^2$. Jump height was determined as the horizontal peak-to-peak displacement of the ankle marker.

Statistics: Group data are presented as mean ± standard deviation (SD). The influence of the seven different acceleration levels was assessed by an analysis of variance for repeated measures, which included contrasts between the values obtained during the 6 acceleration conditions above and below 1*g* and the 1*g* condition.

Results

The main result of the study was that the prerequisites for a reactive jump – the preactivity of the leg extensors and the leg stiffness – could be maintained regardless of the acceleration level, which was also the case for the RFD (see Fig. 4A).

Kinetics: The variation of the acceleration from 0.7g to 1.3g caused a significant, almost linear increase in the peak GRF (+15%, $p < 0.001$) and total impulse during the GCT (+28%, $p < 0.001$), a prolongation of the GCT (+10%, $p < 0.001$) and an increase in the kinetic energy just prior to GC (+24%, $p < 0.001$), as can be seen in Fig. 4B.

Motion analysis: Significant differences in the kinematic parameters as a result of the acceleration variation were observed for the jump height (−21%) and the ankle (+3°) and knee (+3°) excursions during the eccentric phase of the jumps ($p < 0.001$). No significant differences were found for the hip excursion during the eccentric phase and the joint angles at the time of GC, either for the hip or for the knee or ankle (see Fig. 4C and Table 1). There was no heel contact present in any of the jumps.

Muscle activity: During the preactivity phase, the iEMG of the leg extensors showed no significant changes in response to the different acceleration levels, whereas significant phase- and muscle-specific differences were observed for SOL (MLR, LLR), GM (MLR, total), RF (SLR, MLR, total) and most notably BF (all phases, including total), but not for TA and VL. The EMG activity of the 6 leg muscles during 4 phases (PRE, SLR, MLR and LLR) and the total activity during the GCT is shown in Table 2 and illustrated in Fig. 3 for the 0.7*g* and 1.3*g* conditions.

Table 1

Averaged values of all participants of the three joint angles at the time of ground contact (GC), as well as the angular excursion at the hip during the eccentric phase of the jumps. The values in the 1g column are absolute values; negative values indicate that the joint was more flexed than during upright stance, a positive value indicates a more extended joint. The values in the other six columns express the differences of the respective acceleration condition in relation to the 1g values, i.e. a value of −1° indicates that the joint was more flexed than in the 1*g* condition. The standard error of the 40 hops per condition (SE, mean across all subjects and conditions) in the penultimate column illustrates the intra-subject variability. The last column contains the result of the analysis of variance.

	0.7*g*	0.8*g*	0.9*g*	1*g*	1.1*g*	1.2*g*	1.3*g*	SE	*p*-value
Hip angle at GC [°	−1 ± 4	0 ± 3	−1 ± 4	−18 ± 10	−1 ± 3	−1 ± 3	−2 ± 3	0.2	0.16
Knee angle at GC [°]	0 ± 4	−1 ± 3	−1 ± 3	−27 ± 8	−1 ± 3	−1 ± 3	−1 ± 3	0.3	0.72
Ankle angle at GC [°]	0 ± 2	−1 ± 1	0 ± 1	11 ± 4	0 ± 2	0 ± 1	0 ± 2	0.2	0.76
Hip excursion [°]	0 ± 2	0 ± 1	0 ± 1	1 ± 2	0 ± 2	0 ± 1	0 ± 2	0.1	0.75

Table 2

Mean iEMG activity of all participants for the six recorded muscles during 5 phases (PRE 50 ms before GC until GC, SLR 30–60 ms after GC, MLR 60–90 ms, LLR 90–120 ms and total activity from GC until takeoff). To ensure inter-subject comparability, the values in the 1g column are normalized to the iEMG activity from 150 ms before GC until 120 ms after GC during 1g for each subject. For comparison's sake, the values for the other acceleration conditions are expressed in percent of the 1*g* condition. A * symbol denotes a statistically significant difference compared to the 1g condition. The standard error of the 40 hops per condition (SE, mean of the normalized iEMG across all subjects and conditions) in the penultimate column illustrates the intra-subject variability. The last column contains the result of the analysis of variance.

	0.7*g* [%]	0.8*g* [%]	0.9*g* [%]	1g [%]	1.1g [%]	1.2*g* [%]	1.3*g* [%]	SE	*p*-value
PRE SOL	109 ± 28	100 ± 20	94 ± 16	24 ± 10	93 ± 19	104 ± 27	102 ± 39	2%	0.52
PRE GM	102 ± 17	98 ± 12	99 ± 17	37 ± 7	102 ± 10	103 ± 13	102 ± 16	3%	0.34
PRE TA	103 ± 27	107 ± 23	109 ± 32	61 ± 14	111 ± 22	100 ± 12	105 ± 10	5%	0.09
PRE RF	113 ± 47	93 ± 30	98 ± 36	25 ± 8	97 ± 44	106 ± 36	112 ± 58	3%	0.34
PRE VL	109 ± 32	101 ± 19	100 ± 19	22 ± 8	99 ± 19	109 ± 21	111 ± 61	2%	0.14
PRE BF	**97 ± 24**	**96 ± 25**	**99 ± 23**	**33 ± 10**	**114 ± 27***	**119 ± 31***	**125 ± 41***	3%	**< 0.01***
SLR SOL	99 ± 16	99 ± 18	101 ± 16	27 ± 5	105 ± 15	97 ± 17	98 ± 14	9%	0.62
SLR GM	102 ± 7	101 ± 6	102 ± 9	20 ± 1	103 ± 9	103 ± 7	107 ± 9	7%	0.94
SLR TA	101 ± 23	93 ± 15	96 ± 16	12 ± 5	98 ± 17	101 ± 21	94 ± 19	4%	0.39
SLR RF	**95 ± 25**	**100 ± 24**	**107 ± 25**	**25 ± 7**	**106 ± 18**	**123 ± 26***	**126 ± 26***	9%	**< 0.01***
SLR VL	100 ± 24	97 ± 15	104 ± 22	25 ± 3	108 ± 17	102 ± 16	107 ± 21	9%	0.52
SLR BF	**94 ± 14**	**94 ± 12***	**96 ± 11**	**22 ± 5**	**107 ± 25**	**106 ± 15**	**99 ± 15**	8%	**0.04***
MLR SOL	**96 ± 15***	**94 ± 12***	**102 ± 14**	**21 ± 4**	**109 ± 19**	**106 ± 14**	**101 ± 15**	7%	**< 0.01***
MLR GM	**94 ± 21***	**96 ± 11***	**101 ± 16**	**16 ± 3**	**105 ± 19**	**105 ± 12**	**107 ± 20**	6%	**0.02***
MLR TA	91 ± 21	88 ± 15*	98 ± 30	10 ± 4	104 ± 19	102 ± 14	97 ± 25	3%	0.09
MLR RF	**113 ± 54**	**100 ± 23**	**104 ± 22**	**21 ± 7**	**111 ± 31**	**122 ± 29***	**116 ± 3***	9%	**< 0.01***
MLR VL	106 ± 39	96 ± 18	102 ± 23	21 ± 7	111 ± 26	109 ± 21	110 ± 27	8%	0.31
MLR BF	**88 ± 25***	**97 ± 24**	**93 ± 20**	**16 ± 5**	**108 ± 30**	**123 ± 32***	**120 ± 34***	6%	**< 0.001***
LLR SOL	**82 ± 27***	**86 ± 22***	**103 ± 43**	**10 ± 6**	**125 ± 55**	**118 ± 26***	**112 ± 29***	4%	**< 0.01***
LLR GM	86 ± 51	90 ± 28	109 ± 66	9 ± 5	129 ± 47	127 ± 32*	121 ± 37*	3%	0.07
LLR TA	116 ± 85	97 ± 52	96 ± 37	7 ± 3	105 ± 37	116 ± 25	104 ± 35	5%	0.51
LLR RF	113 ± 40	98 ± 28	96 ± 29	12 ± 7	109 ± 27	116 ± ± 37	116 ± 46	6%	0.79
LLR VL	105 ± 37	91 ± 28	97 ± 27	10 ± 5	118 ± 43	107 ± 37	126 ± 67	4%	0.66
LLR BF	**101 ± 38**	**110 ± 58**	**115 ± 51**	**9 ± 4**	**129 ± 69**	**138 ± 4***	**145 ± 75***	4%	**< 0.01***
GCT SOL	96 ± 12	96 ± 10	100 ± 12	81 ± 12	108 ± 17	106 ± 11	103 ± 10	4%	0.06
GCT GM	**97 ± 13**	**97 ± 10**	**101 ± 10**	**67 ± 8**	**106 ± 12**	**110 ± 8***	**109 ± 10***	4%	**0.03***
GCT TA	92 ± 12*	90 ± 13	96 ± 20	63 ± 22	99 ± 10	108 ± 15*	108 ± 22*	5%	0.07
GCT RF	**98 ± 21**	**96 ± 17**	**103 ± 18**	**92 ± 17**	**109 ± 19**	**120 ± 16***	**120 ± 16***	5%	**< 0.01***
GCT VL	100 ± 18	96 ± 12	102 ± 18	89 ± 13	111 ± 22	106 ± 16	109 ± 19	4%	0.48
GCT BF	**94 ± 19***	**95 ± 16***	**98 ± 17**	**78 ± 12**	**108 ± 23**	**120 ± 20***	**119 ± 26***	4%	**< 0.001***

Discussion

The main finding of this study is that regardless of the acceleration level, the subjects were able to maintain a high preactivity of the leg extensors and also a high leg stiffness, enabling them to perform reactive hops with accelerations ranging from 0.7 to 1.3g.

The increase of the acceleration led to an increase of the kinetic energy just prior to GC, a parameter that is considered to be one of the most important factors determining the load that the jumper is subjected to (Bubeck, 2002). A physical point of view, it suggests itself that the increase of the GRF, the impulse and the angular excursions in the knee and ankle, as well as the prolongation of the GCT were a result of this increasing kinetic energy, similar to a spring which is compressed more and for a longer time when more energy is supplied.

In line with the biomechanical adjustments, the modulation of the acceleration was accompanied by muscle and phase-specific neuronal adjustments: (a) the preactivity of the leg extensors remained constant, (b) amongst the EMG phases – SLR, MLR, LLR – where reflex activity is expected, MLR was the most affected by the acceleration level. The preactivity determines the setting of the tendomuscular system at the time of GC, and the preactivity of the leg extensors in particular has been argued to be the major determinant of leg stiffness (Gollhofer and Kyröläinen, 1991). Consequently, a constantly high preactivity of the leg extensors seems necessary to maintain a high leg stiffness. The fact that the phase of the SLR was hardly affected by the modulation of the acceleration could be explained by the fact that the SLR is mainly sensitive to the stretch velocity (Gottlieb and Agarwal, 1979), which can be assumed to have been rather constant since the leg stiffness and the RFD did not change with increasing acceleration. For the increase in EMG activity observed in four of the six muscles during the phase of the MLR, two explanations are proposed: since the MLR was shown to be sensitive to changes in length rather than velocity (af Klint et al., 2010; Gray et al., 2001), the increase could be due to changes in the stretch amplitude, which potentially increased due to the significantly increased ankle and knee excursions. The other possible explanation is also applicable to the LLR: it has been shown that both the MLR and the LLR can be modulated by supraspinal structures (Diener et al., 1985; Mrachacz-Kersting et al., 2006). Therefore, it is possible that in addition to increases in spinal reflex activity caused by increased stretch amplitudes, supraspinal centers were responsible for the increase in EMG activity during the phases of the MLR and LLR, either via transcortical reflexes or an adjustment of the preprogrammed central motor program. Examining the EMG activity, it is noteworthy that the antagonistic muscle BF was the only muscle to show a significantly increased activity in each of the analyzed phases. Such a coactivation of antagonistic muscles when joint stabilization is required has been well documented for a variety of tasks (Aagaard et al., 2000; van Dieën et al., 2003; Zakotnik et al., 2006). Consequently, it is suggested that the observed coactivation of BF is a mechanism that serves to stabilize the joints in order to cope with the increasing load induced by the increasing acceleration.

The observation that the participants could maintain a constantly high preactivity and leg stiffness underlines the ability of the neuromuscular system to maintain a particular movement pattern in spite of the modulated acceleration. The reason why this was not possible in previous studies cannot be pinpointed, but there are several possible explanations, mostly related to the methodological approaches previous studies have used. For example, the system employed by Bubeck and Gollhofer (2002,2001) achieved the modulation of the acceleration by adding or subtracting weights from a weight stack attached to the lying jumper. The weight stack's mass considerably increased the jumper's total inertia and thus led to a huge increase in energy turnover during ground contact. Other systems could circumvent the inertia problem, but had to sacrifice the uniformity of the acceleration level: both with the system used by Gollhofer and Kyröläinen (1991) and with the one used by Avela et al. (1994), the effective acceleration was achieved via counterweights that were connected during the fall phase of the jump only, i.e., at the moment of GC there was a sudden system-inherent change in acceleration, which apparently was something the neuromuscular system had problems coping with.

Furthermore, when comparing the results of the present study to those of previous studies, it is important to keep in mind that in contrast to the single drop jumps used in most previous studies, the present study used repeated hops. Hops were chosen because a high number of repetitions are possible in a short time period, which leads to a reliable mean and reduces potential confounding effects of fatigue. However, hops are a self-constrained movement pattern, since each hop has an effect on the next one: for instance, the falling height is determined by the jump height of the previous hop, whereas it is preset in the case of drop jumps. Thus, the present study shows that accelerations above or below 1g do not present a fundamental obstacle, but it cannot establish an exact range for the factors that limit the possibility to perform reactive jumps, i.e., falling height, acceleration and kinetic energy.

Another limitation of the present study is that the participants' experience with reactive jumps differed, depending on the sport they practiced. This is probably the reason for the relatively high inter-subject variation observed in response to the different acceleration levels. Lower variation would have allowed an even more detailed analysis of the acceleration's influence, particularly when comparing the values recorded during the acceleration levels close to 1g to the values obtained during 1g.

Lastly, it is important to carefully consider which instruction is given to the participant regarding the way the jumps should be performed: Arampatzis et al. (2001) showed that the jumping strategy and in particular the leg stiffness can be modified by giving different instructions . Therefore, it cannot be excluded that some of the differences between the results of the present study and previous studies were due to different instructions, lacking control of the adherence to the instructions or even a lack of any specific instructions at all. However, some of the previous studies used similar instructions – "jump reactively with only little bending in the knee" (Gollhofer and Kyröläinen, 1991) – as in the present study – "jump as stiff as possible" – which means that the bulk of the differences was most likely due to the different approaches used to modulate the acceleration.

A practical application of the SJS could be as a training device for rehabilitation purposes: athletes recovering from injuries who are not yet able to tolerate the high loads associated with their regular plyometric training could still perform a specific plyometric training, but with reduced acceleration and therefore also reduced load. Another application could be as a training device for athletes looking for a nongeneric way to train with increased loads, which could be achieved by using accelerations above 1g.

In summary, it is concluded that by maintaining a high preactivity of the leg extensors and adjusting the muscle activity in the later phases, the subjects were able to maintain a high leg stiffness and perform reactive hops throughout the acceleration range of 0.7–1.3g. It is therefore argued that the neuromuscular system seems to be able to cope with different acceleration levels, enabling the subjects to maintain a specific parameter – in this case the leg stiffness – despite the numerous changes accompanying the modulation of the acceleration.

Conflict of interest statement

The authors did not have any financial and personal relationships with other people or organizations that could inappropriately influence the study.

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