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# Influence of gravity compensation on muscle activity during reach and retrieval in healthy elderly

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**Abstract**

*Introduction:* Arm support like gravity compensation may improve arm movements during stroke rehabilitation. It is unknown how gravity compensation affects muscle activation patterns during reach and retrieval movements. Since muscle activity during reach is represented by a component varying with movement velocity and a component supposedly counteracting gravity, we hypothesized that gravity compensation decreases the amplitude of muscle activity, but does not affect the pattern. To examine this, we compared muscle activity during well defined movements with and without gravity compensation in healthy elderly.

*Methods:* Ten subjects performed reach and retrieval movements with and without gravity compensation. Muscle activity of biceps, triceps, anterior, middle and posterior parts of deltoid and upper trapezius was compared between the two conditions.

*Results:* The level of muscle activity was lower with gravity compensation in all muscles, reaching significance in biceps, anterior deltoid and trapezius ($p \leqslant 0.026$). The muscle activation pattern did not differ between movements with and without gravity compensation ($p \geqslant 0.662$).

*Discussion:* Gravity compensation only influenced the level of muscle activity but not the muscle activation pattern in terms of timing. Future studies should examine if the influence of gravity compensation is comparable for stroke patients. This may stimulate early and intensive training during rehabilitation.


*Keywords:* Muscle activity; Upper extremity; Gravity compensation; Healthy persons; Elderly

## 1. Introduction

A stroke can result in disturbed sensory and motor functions of contra-lateral body parts (i.e., hemiparesis), along with communicative and cognitive disorders. With respect to the upper extremity, impaired arm and hand function causes limitations in activities of daily living for the majority of stroke patients; only 5–20% of the patients experience full recovery, measured at 6 months post-stroke (Kwakkel et al., 2003). Recovery of arm function after stroke can be stimulated by an intensive program of exercise therapy (i.e., frequent and/or extended sessions) that involves active movement of the affected arm (Feys et al., 1998; Kwakkel et al., 1999; Barreca et al., 2003).

In clinical practice, active arm movements are frequently facilitated by supporting the weight of the arm, for example by suspending the arm from a frame overhead. Recently, technological innovations such as robotic devices are also applied to stimulate recovery of arm function after stroke (Prange et al., 2006). These devices often include arm support, to compensate for the effect of gravity on the arm (Johnson, 2006). Research on arm support for stroke patients, using a frictionless air slide above a table, showed an increase in maximal voluntary torques of isometric elbow extension when the hemiparetic arm was supported

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against gravity (Beer et al., 1999). A subsequent study showed that the active range of elbow extension increased with arm support during reaching movements (Beer et al., 2004). This indicates that using arm support may be beneficial to improve arm movements in post-stroke rehabilitation, since patients do not have to use their remaining muscle strength and coordination to lift and hold their arm themselves.

Although for stroke patients some information about the influence of gravity compensation on kinetic and kinematic characteristics of reaching movements (like muscle torques, range of motion and movement velocity) is available, only little is known about the way gravity compensation influences muscle activity.

Research on the kinematics of goal-directed arm movements in micro-gravity during parabolic flights has indicated that gravity is implemented in internal models used in planning of movements by the central nervous system (Smetanin and Popov, 1997; Papaxanthis et al., 1998; Pozzo et al., 1998). Therefore, compensation of the influence of gravity on the arm may affect the muscle activation patterns of reaching movements. A study into differences between arm movements in different gravitational loads showed that different patterns of muscle activation occurred during movements with gravity and against gravity (Virji-Babul et al., 1994).

On the other hand, muscle activity signals during forward-directed reach were found to be determined by two principal components, of which the contribution to the muscle activity signal depended on the weighting factor of each component (Flanders and Herrmann, 1992). One 'phasic' component varied with movement velocity, the other 'tonic' component was unrelated to movement velocity and was assumed to be involved in counteracting gravity during reaching. The scaling of the phasic component with movement velocity was in line with the recorded muscle activity during reaching with different speeds. However, it was not verified whether the tonic component did relate to counteracting gravity (Flanders and Herrmann, 1992).

To examine this, the present study aimed to investigate the influence of gravity compensation on muscle activation patterns during reach and retrieval movements performed by healthy elderly. Based on the study by Flanders and Herrmann (1992), we would expect that when the influence of gravity on the arm is compensated during reaching, the amplitude of muscle activity is affected, but the temporal pattern remains unchanged.

## 2. Methods

### 2.1. Subjects

Ten healthy persons participated in this study, after providing written informed consent. Inclusion criteria were that the subjects were over 50 years of age and had no known history of neuromuscular, orthopedic or rheumatologic disorders of the upper extremity. The study was approved by the local medical ethics committee.

### 2.2. Apparatus

A mechanical, passive device was developed to counteract the influence of gravity on the upper extremity (named Freebal, derived from 'balanced freedom'). The device consisted of two independent springs, connected to pliable joint braces at the elbow and wrist via ropes and pulleys overhead (Fig. 1). This system provided a constant and individually adjustable amount of gravity compensation throughout the entire workspace of approximately 1 $m^3$, enabling three dimensional movements. A more detailed description of the Freebal is published elsewhere (Stienen et al., 2006, 2007).

### 2.3. Procedure

The subjects performed two series of movements with the dominant arm, once with and once without gravity compensation. To reduce the potential effect of learning or adaptation, the

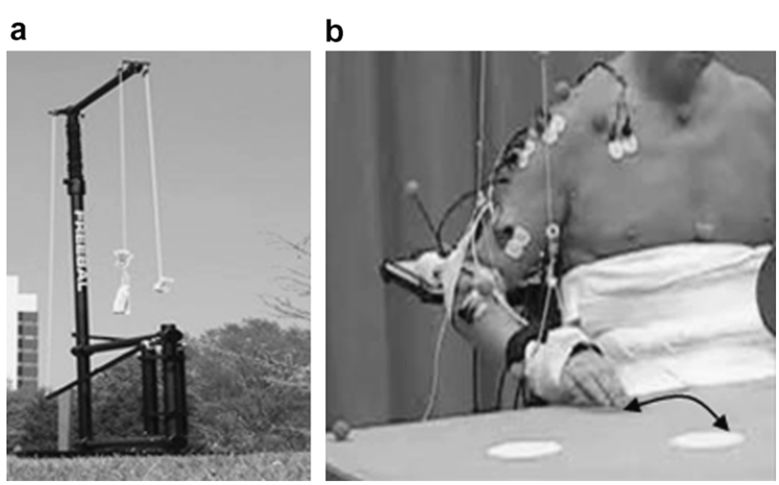


Fig. 1. Apparatus for gravity compensation: (a) the Freebal device in full view; (b) close-up of the arm attached to the wrist and elbow straps of Freebal, with the hand on the starting dot and the arm in the initial posture; the arrow indicates the required movement from starting to target dot and back; also visible are EMG electrodes and reflective markers (see Section 2.4 for details).

subjects were assigned randomly to one of two sequences, either performing the movements first with and subsequently without gravity compensation (group A), or in reversed order (group B).

The subjects were seated at an in height adjustable table and secured to the chair with straps to limit compensational trunk movements. Subjects sat with the upper arm parallel to the trunk, the elbow flexed 90°, the forearm in pronation and the wrist fixated in a neutral position midway between flexion/extension and radial-/ulnar-abduction by a splint. In this position the hand was placed on the starting dot (Fig. 1b). The target dot was placed 35 cm from the starting dot, so that the subjects reached forward in the sagittal plane using both the shoulder and the elbow. Both dots had a diameter of 10 cm.

The subjects performed repeated multi-joint reach and retrieval movements during 30 s, alternating between starting and target dot. The subjects were instructed to move at their own comfortable pace and touch the centre of the dots, placing more focus on accuracy than on velocity.

### 2.4. Measurements and data recording

#### 2.4.1. EMG recordings

Bi-polar surface electromyography (EMG) was recorded using disc-shaped wet-gel Neuroline Ag/AgCl-electrodes (type 72000-S; Medicotest A/S, Ølstykke, Denmark) with a recording area of 95 $mm^2$ and an inter-electrode distance of 2 cm. Electrodes were placed over six superficial muscles, according to guidelines of the SENIAM project: long head of biceps (BIC); long head of triceps (TRI); anterior (DA), middle (DM), posterior (DP) part of deltoid; upper part of trapezius (TRA). (Hermens et al., 1999) After differential amplification using a K-Lab amplifier (K-Lab, Haarlem, the Netherlands; input impedance 10 GΩ, common mode rejection ratio >110 dB, input voltage noise <2 μV, gain 18750), the EMG signals were high-pass filtered (third-order Butterworth filter, cut-off frequency 20 Hz) and digitized by a 12-bit analog-to-digital converter with a sample rate of 1000 Hz. These EMG signals were then band-pass filtered (second-order zero phase shift Butterworth, cut-off frequencies 20–400 Hz) and converted to smooth rectified EMG (SRE) signals (using a low-pass second-order zero phase shift Butterworth filter at 25 Hz for smoothing) for each muscle per subject.

#### 2.4.2. Kinematic data

Changes in positions of arm segments were recorded using an infrared 3D-motion analysis system (VICON 370: six cameras, sample rate 50 Hz; Oxford Metrics Ltd, Oxford, United Kingdom). Reflective markers were placed on 10 bony landmarks on the arm and trunk and converted to joint angles according to guidelines of the International Society of Biomechanics (ISB) (Wu et al., 2005).

The elbow joint angle was specified as the angle between humerus and forearm (maximal elbow flexion is 0°, maximal elbow extension is 180°). The shoulder joint orientation was described using two angles (see Fig. 2 for illustration). First, the plane of elevation was defined as the angle of the humerus with a virtual line through both shoulders, viewed in the transversal plane (outward/lateral is 0°, arm extended forward is 90°). In other words, it represented the angle of the projection of the upper arm on the horizontal plane. Second, the angle of elevation was the angle between humerus and trunk in the plane of elevation (consisting of the vertical plane through the upper arm), irrespective of the orientation of the humerus in the transversal plane (humerus parallel with trunk is 0°, humerus parallel with horizontal is 90°). These angles were low-pass filtered at 20 Hz with a second-order zero phase shift Butterworth filter and linearly interpolated from 50 to 1000 Hz to match the sample rate of the EMG recordings.

#### 2.4.3. Synchronous display of EMG and kinematics

Custom written software using the Labview environment (National Instruments, Austin, TX, United States) was used to analyse SRE signals in relation to joint angles. Reversals in movement direction were defined at minimum and maximum elbow joint angles (hand on starting and target dot, respectively), determined by zero-crossings of the elbow angular velocity. A reach (minimal to maximal elbow angle) and subsequent retrieval (maximal to minimal elbow angle) movement constituted a movement cycle (see Fig. 3).

### 2.5. Data analysis

The EMG and joint angle data were visually inspected for missing data or recording errors. The SRE-values and joint angle trajectories were averaged over all repeated movement cycles of one 30 s series of movements to obtain an average muscle activation pattern (MAP), which was normalized to the average duration of all repeated movement cycles. The reach and retrieval trajectories were each divided in three movement parts: part (I) initiation of movement, defined by 1–20% of a trajectory; part (II) steady state movement during the middle part of a trajectory, from 31% to 70%; part (III) termination of movement, consisting of 81–100% (see Fig. 3). Per subject, the SRE-values of each MAP were averaged within each movement part as a measure of level of muscle activity.

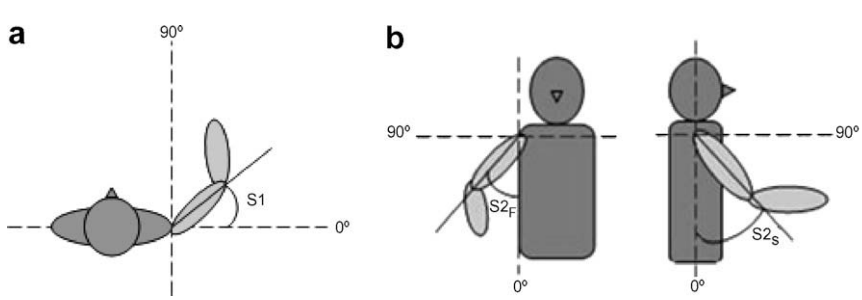


Fig. 2. Definition of shoulder angles: (a) S1: plane of elevation, angle of the humerus with a virtual line through both shoulders; (b) S2: angle of elevation, angle between humerus and trunk in the plane of elevation (consisting of the vertical plane through the upper arm), represented in frontal view ($S2_F$) and sagittal view ($S2_S$).

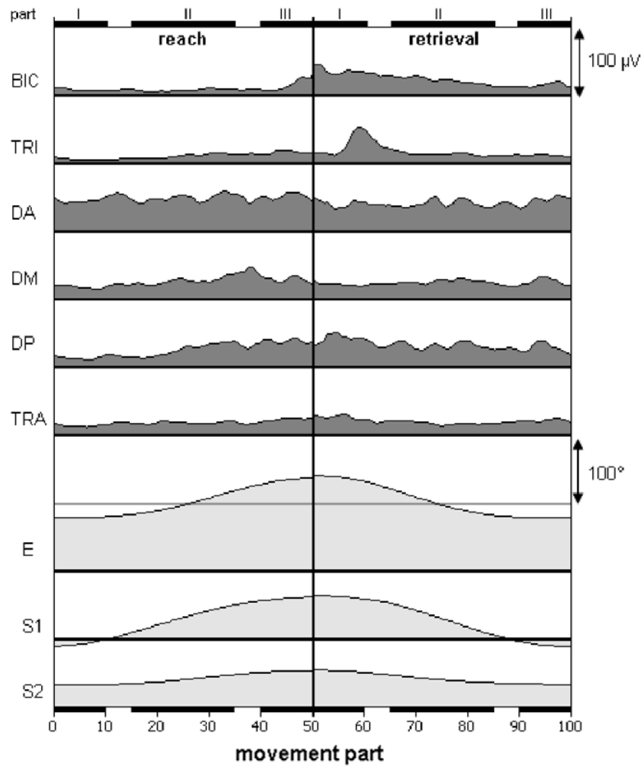


Fig. 3. Muscle activation pattern (MAP) of six muscles and corresponding joint angles during movement without gravity compensation (time in % of averaged cycle duration) of one subject, including definition of movement parts by black bars. Abbreviations: part I = initiation, II = steady state, III = termination of movement; BIC = long head of biceps; TRI = long head of triceps; DA = anterior deltoid; DM = medial deltoid; DP = posterior deltoid; TRA = upper trapezius; E = elbow flexion/extension angle; S1 = shoulder plane of elevation; S2 = shoulder angle of elevation.

To quantify movements, several parameters were derived from kinematic data: movement time was defined as the movement cycle duration averaged over all movement cycles per subject (in ms); movement symmetry was defined as the relative duration of reach with respect to the average movement cycle (in %); and joint excursions of elbow and shoulder angles were defined as the difference between minimal and maximal joint angles (in °), averaged over all movement cycles.

### 2.6. Statistical analysis

Initially, SRE-values were transformed to LN(SRE)-values to ensure normal distribution of residuals. To detect potential effects of learning or adaptation, muscle activity and movement performance parameters of the two groups performing movements with and without gravity compensation in reversed order (A and B) were compared using a *t*-test for independent samples in the case of normally distributed parameters and a non-parametric equivalent, the Kolmogorov–Smirnov test, for parameters deviating from the normal distribution. Furthermore, either paired-samples *t*-tests or the non-parametric equivalent, Wilcoxon signed ranks tests, were used to examine differences in movement execution parameters (movement time, movement symmetry and joint excursions) due to gravity compensation. For all these tests the significance level was 0.05.

Linear mixed modeling techniques (i.e., multilevel linear modeling, random coefficient analysis) were applied to test differences in SRE-values due to gravity compensation (two-level factor ‘fz-comp’) in each muscle (six-level factor ‘muscle’) per movement part (three-level factor ‘part’), separately for reach and retrieval. To account for the correlation of repeated measurements within one subject, the factor ‘subject’ was included as a random factor. The factors ‘fz-comp’, ‘muscle’ and ‘part’ were treated as fixed effects, because the effect of gravity compensation on muscle activity was similar for all subjects. The two-way interactions ‘muscle × part’, ‘fz-comp × muscle’ and ‘fz-comp × part’ were included to examine the change of muscle activity across movement parts per muscle, the difference of the influence of gravity compensation on the level of muscle activity between muscles and the influence of gravity compensation on the change of muscle activity across movement parts, respectively. For all significant effects and interactions post-hoc tests (Sidak adjustment) were performed. The level of significance was defined as 0.05.

## 3. Results

### 3.1. Subjects

Data of all 10 subjects were used for analysis, of which average values of physical characteristics are presented in Table 1. Regarding the order of conditions, seven subjects moved first with and next without gravity compensation (group A), while three subjects performed the test in the reversed order (group B). There were no significant differences in physical characteristics, movement execution parameters and SRE-values between the two groups, so data of all subjects were pooled in subsequent analyses.

### 3.2. Movement execution

Movement execution parameters averaged over all subjects are presented in Table 2. On average, 19 (±8) movement cycles per subject were analyzed. Comparison of movement execution with and without gravity compensation showed that movement time and movement symmetry were not significantly different ($p = 0.224$, $p = 0.160$, respectively). The joint excursions of elbow and shoulder plane were also comparable for both conditions ($p = 0.376$ and $p = 0.102$, respectively), but the excursion of shoulder elevation was somewhat smaller with gravity

Table 1
Physical characteristics of 10 healthy elderly

| | Subjects ($n = 10$) |
|---|---|
| Sex (male/female)[a] | 5/5 |
| Arm dominance (right/left)[a] | 10/0 |
| Age (years)[b] | 65.9 (±6.4) |
| Height (m)[b] | 1.74 (±0.12) |
| Weight (kg)[b] | 77.7 (±13.6) |
| Body mass index (kg/m$^2$)[b] | 25.7 (±2.9) |

[a] Absolute numbers.
[b] Mean ± SD.

Table 2
Movement execution parameters (mean ± SD) and corresponding inter-subject coefficients of variation (COV = SD/mean) for movements without and with gravity compensation of 10 subjects

| | Without compensation | With compensation |
|---|---|---|
| Elbow excursion (°) | 60.4 (±4.4) | 61.2 (±6.6) |
| *Inter-subject COV* | 0.07 | 0.11 |
| Shoulder plane excursion (°) | 81.6 (±11.7) | 78.2 (±11.9) |
| *Inter-subject COV* | 0.14 | 0.15 |
| Shoulder elevation excursion (°)* | 23.2 (±9.5) | 18.8 (±6.8) |
| *Inter-subject COV* | 0.41 | 0.36 |
| Movement time (ms) | 1327 (±383) | 1491 (±211) |
| *Inter-subject COV* | 0.29 | 0.14 |
| Movement symmetry (%) | 56.8 (±3.7) | 58.8 (±1.7) |
| *Inter-subject COV* | 0.07 | 0.03 |

* Significant difference between movement with and without gravity compensation ($p < 0.05$).

compensation ($p = 0.013$). However, the difference in shoulder elevation excursion was not even 5°, which was not regarded as a substantial difference in movement execution.

### 3.3. Generic aspects of muscle activation

In Fig. 3 a typical example of a MAP of movement without gravity compensation is displayed, which revealed some features in this subject. Most muscles were active throughout the movement, without periods of relative rest. An exception is TRI, which had very low levels of activity during reach and retrieval (except at the start of retrieval in this particular subject).

When comparing MAPs of movements without gravity compensation across subjects (Fig. 4), several generic aspects could be identified. Most muscles had persistent levels of muscle activity throughout the reach and retrieval movements, except that almost all subjects showed rather low levels of TRI activity. Furthermore, most muscles showed increasing SRE-values from movement initiation to termination during reach, as supported by a significant main effect of 'part' ($p = 0.000$). This general increase in SRE-values differed between muscles, as revealed by a significant interaction of 'muscle × part' ($p = 0.000$). Post-hoc analyses (represented by asterisks in Fig. 4) showed increases in SRE-values from steady state to termination in BIC ($p = 0.008$) and from initiation to steady state in TRI ($p = 0.000$) and in DP ($p = 0.001$), while in TRA SRE-values decreased from initiation to steady state ($p = 0.036$) and subsequently increased ($p = 0.033$). During retrieval, a general decrease in muscle activity was observed towards movement termination (main effect of 'part'; $p = 0.009$), which was not significantly different between muscles ('muscle × part' $p = 0.106$).

When integrating the qualitative and quantitative findings, generic aspects of muscle activation patterns could be identified for reach and retrieval without gravity compensation. BIC was active to lift and hold the lower arm above the table and aid in anteflexion of the shoulder.

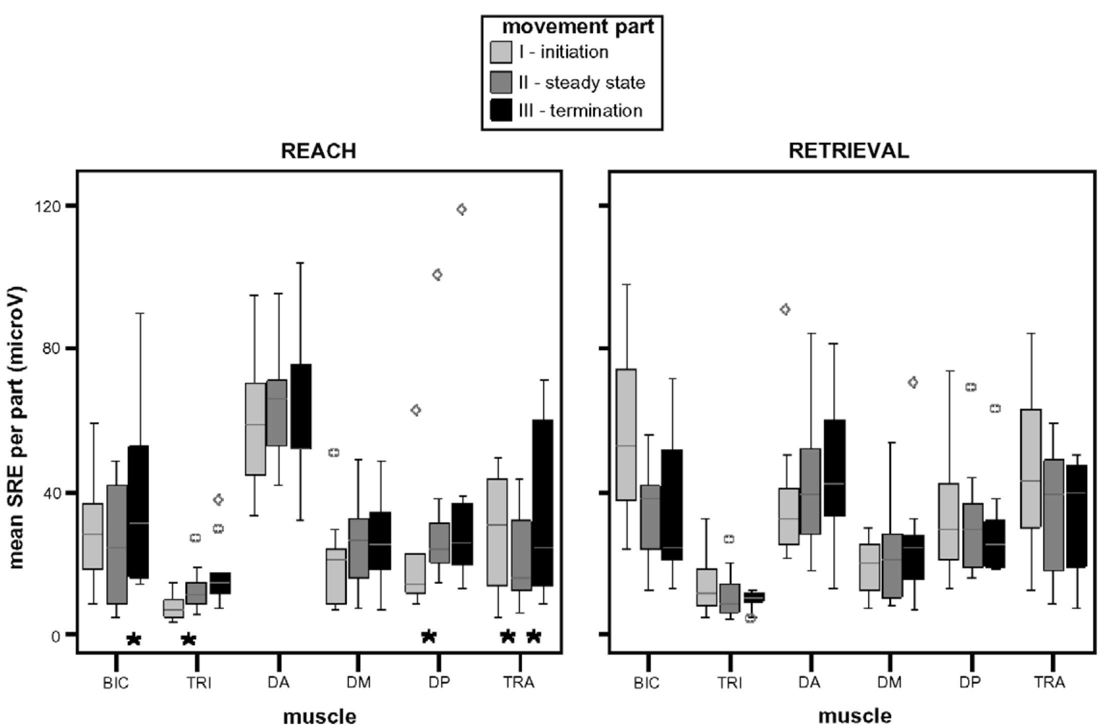


Fig. 4. Median and interquartile ranges of SRE-values of 10 subjects per movement part for each muscle during movement without gravity compensation, for reach and retrieval separately. Muscle activity significantly increased during reach and significantly decreased during retrieval; muscle-specific changes in SRE-values from one part to next during reach are represented by asterisks (*$p < 0.05$). Circles (○) represent outliers (deviation of <1.5 times interquartile range) and diamonds (◇) represent extremes (deviation of <3 times interquartile range).

Muscle activity of TRI was very low, sometimes nearly absent, contributing to extension of the elbow towards the target. DA and DM were active to maintain a certain degree of shoulder abduction, to anteflex the shoulder during reach and to decelerate retroflexion during retrieval. Activity of DP decelerated anteflexion during reach and retroflexed the shoulder during retrieval. TRA was active to elevate the arm at the start of reach and retrieval and to position the scapula appropriately during the movements.

### 3.4. Influence of gravity compensation on muscle activation

Comparison of MAPs with and without gravity compensation (Fig. 5) showed a reduction in the level of activity of most muscles with gravity compensation, as underlined by a significant overall effect of 'fz-comp' ($p = 0.000$) for both reach and retrieval. During retrieval, the influence of gravity compensation differed between muscles as indicated by a significant interaction 'fz-comp-muscle' ($p = 0.007$). Post-hoc differences (in Fig. 5 accentuated by asterisks) between movements with and without gravity compensation were found in BIC ($p = 0.000$), DA ($p = 0.000$) and TRA ($p = 0.026$). These same muscles also showed significant post-hoc differences between the two conditions during reach ($p < 0.007$), but these differences were smaller than during retrieval, resulting in a non-significant interaction of 'fz-comp $\times$ muscle' ($p = 0.252$) during reach.

Despite differences in the level of activity, the change of muscle activity across movement parts (Fig. 6) was similar for movements with and without gravity compensation for most muscles, as supported by a non-significant interaction 'fz-comp $\times$ part' for both reach and retrieval ($p = 0.662$ and $p = 0.826$, respectively). Although TRI and DP displayed a deviating pattern with a slightly larger level of muscle activity at the end of reach and start of retrieval with gravity compensation, these differences were too small to result in a significant effect.

## 4. Discussion

The aim of the present study with healthy elderly was to investigate the influence of gravity compensation on muscle activation patterns during reach and retrieval movements. A common aspect of muscle activity during movements without gravity compensation across all subjects was that most muscles show persistent activity during the entire reach and retrieval trajectories. As an exception, activity was very low in TRI. In general, the persistent activity increases during reach and decreases during retrieval. A plausible explanation is that larger torques act on the extended arm at the target dot then when the arm is close to the trunk at the starting dot, requiring a higher level of muscle activity at the end of reach and start of retrieval.

The generic aspect of persistent activity during reach and retrieval corresponds largely with the findings of Sabatini et al., who investigated 'natural' unsupported reaching movements with a similar arm orientation (shoulder abduction of approximately 45°, hand at table height) as the present study (Sabatini, 2002). They found that all recorded muscles (BIC, TRI, DA, DP and TRA) were active throughout the movement, with additional activity on top of this persistent activity in BIC, DP and TRA.

In contrast, several other studies described an alternating pattern of BIC/TRI and DA/DP activity during reach

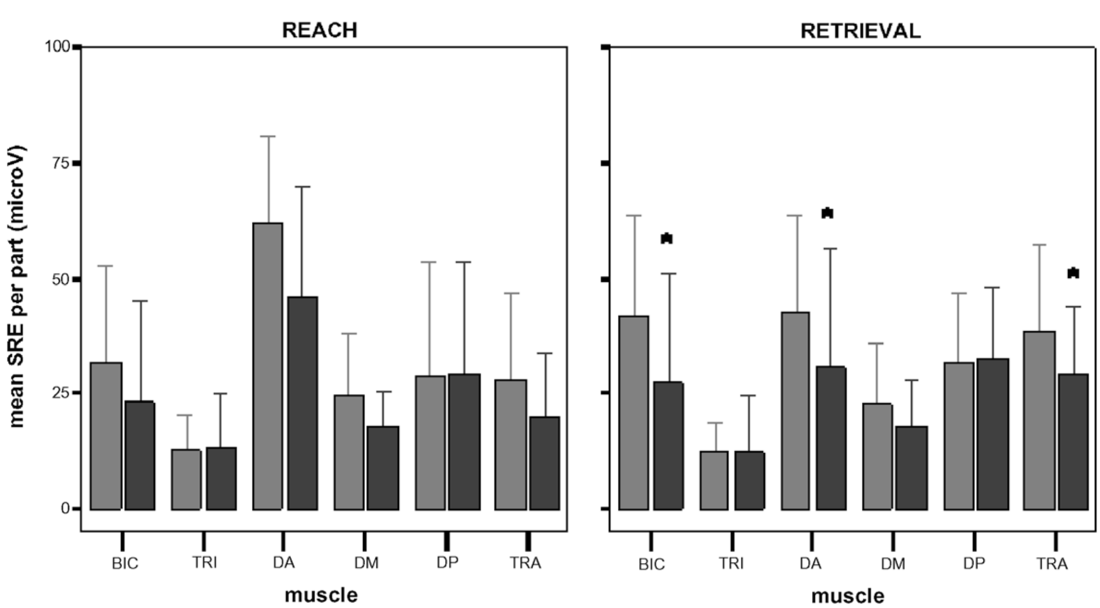


Fig. 5. Comparison of mean (±SD) SRE-values, averaged over all three parts per muscle, between movement without (light bars) and with (dark bars) gravity compensation, for reach and retrieval separately ($n = 10$); asterisks represent significant differences ($p < 0.05$).

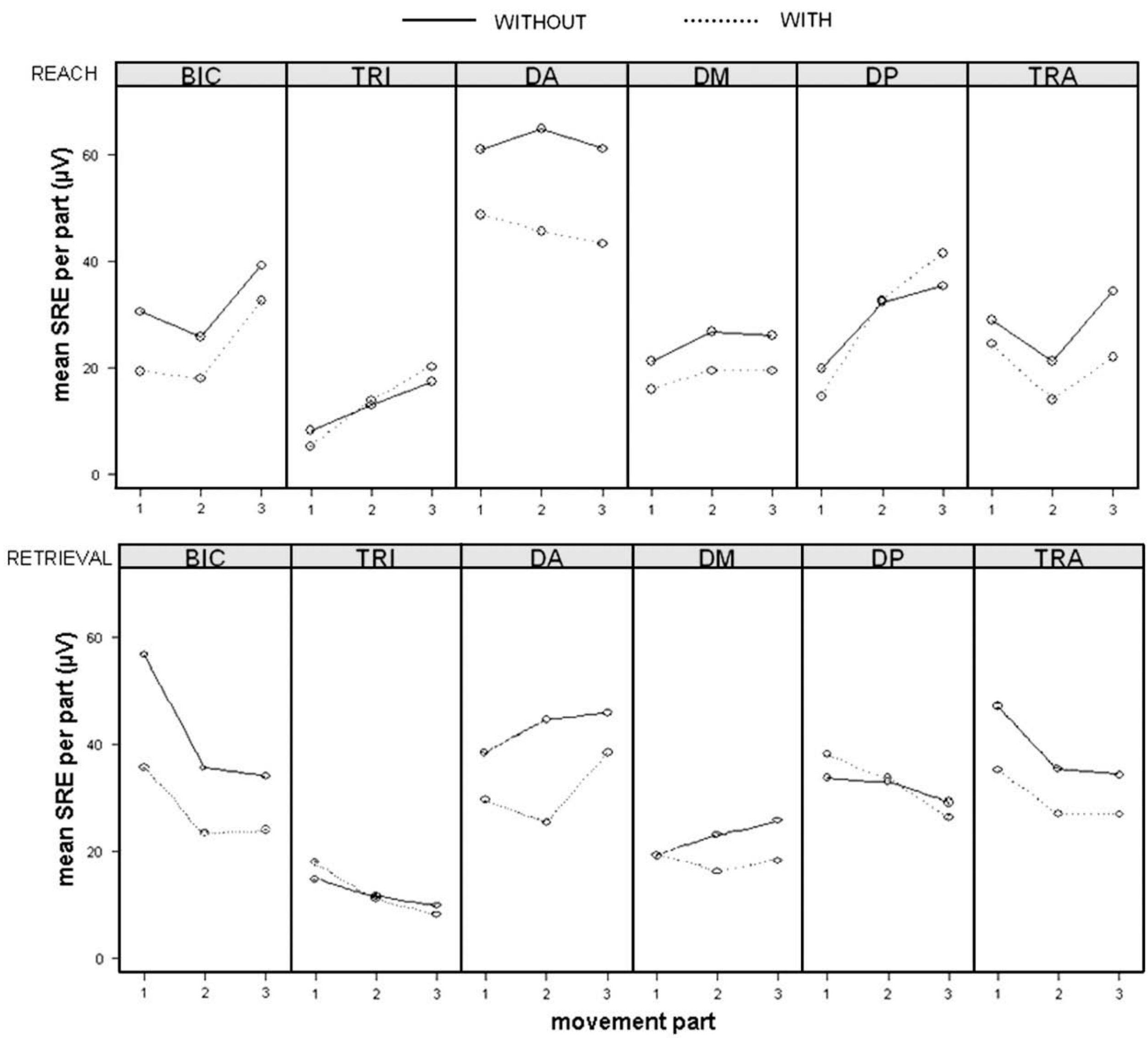


Fig. 6. Changes in mean SRE-values across movement parts compared for movements without (solid lines) and with (dotted lines) gravity compensation of 10 subjects per muscle and for reach and retrieval separately.

and retrieval with arm support, interspersed with periods of relative rest (Karst and Hasan, 1991; Dounskaia et al., 2002; Gribble et al., 2003). This discrepancy is probably related to the orientation of the arm during movements. In the present study the targets were at table height, requiring a certain amount of arm elevation against gravity to reach the target, while in the reported literature both the targets and the arm were in the horizontal plane at shoulder height (approximately 90° shoulder elevation, with arm support) and no movement against gravity was necessary. A study examining timing of muscle activity for movements in different directions reported that single-joint elbow movements against gravity resulted in periods of simultaneous activity of agonist and antagonist (i.e. co-activation), while movements in the horizontal plane produced an alternating pattern of agonist–antagonist activity (Karst and Hasan, 1991). Thus, phasic activity may have been largely masked by a higher level of persistent activity during movements in the present study, requiring movement against gravity.

Besides arm orientation, other differences between the present study and the reported studies (Karst and Hasan, 1991; Dounskaia et al., 2002; Gribble et al., 2003) may also play a role. For example, in the present study movement velocity was lower, especially with respect to the study by Gribble et al. where subjects were instructed to make movements as fast as possible (Gribble et al., 2003). Several studies have indicated that a higher movement velocity is related to a higher intensity of muscle activity (Buneo et al., 1994; Gabriel, 1997) and to the presence of a typical tri-phasic pattern of agonist/antagonist activity (Brown and Gilleard, 1991; Berardelli et al., 1996). In addition, the present study included older subjects than several other studies (Dounskaia et al., 2002; Gribble et al., 2003), whom have been reported to have a lower depth of muscle activity modulation, meaning that in elderly the range from minimal to maximal values of generated muscle activity is smaller (Ketcham et al., 2004).

Comparison of muscle activation patterns during reach and retrieval movements with and without gravity compensation showed that the level of muscle activity was lower with gravity compensation, while no substantial differences in movement execution between the two conditions were found. This difference was observed specifically in the muscles that counteract gravity during forward reach and retrieval movements: BIC, DA and TRA. The change of muscle activity across movement parts did not differ between the two conditions.

The observed decrease in the level of muscle activity without differences in the pattern of muscle activation confirmed our hypothesis, based on a study by Flanders and Herrmann (1992) suggesting that tonic muscle activity was related to counteracting gravity while phasic activity was not. These findings indicate that performing movements without the effect of gravity on the arm only reduced the level of muscle activity and did not affect the pattern of muscle activation in terms of timing.

Another study investigating a basic application of gravity compensation, manual support of gross arm movements in healthy persons, compared an unspecified parameter of muscle activity during supported arm elevations with unsupported movements. This study reported less muscle activity of the serratus anterior and the pectoralis major muscles during supported movements (Hiengkaew et al., 2003). Despite the use of other movements and muscles and a lack of specificity on analysis of muscle activity, this outcome is along the same lines of the influence of gravity compensation observed in the present study.

Remarkably, the application of gravity compensation did not lead to unmasking of phasic components, but a certain degree of tonic activity remained during movements with gravity compensation. This may be related to the many degrees of freedom that characterize upper extremity movements and the interaction torques acting on a joint due to movements of an adjacent joint. Simultaneous application of both agonist and antagonist activity can stabilize the joints and control the interaction torques during movements (Dounskaia et al., 2002; Gribble et al., 2003). This co-activation presents as 'tonic' activity during arm movements, even when the arm is supported.

The findings of the present study show that healthy elderly performed similar movements with a lower level of muscle activity, while the change of muscle activity across movement parts did not differ during movements with gravity compensation. This indicates that when performing movements while the arm is supported against gravity only the level of muscle activity is influenced, but the pattern of muscle activation in terms of timing is not affected, compared to movements without support.

Moreover, this implies that, if the influence of gravity compensation on movements of stroke patients is comparable to healthy elderly, gravity compensation may be able to facilitate arm movements during post-stroke rehabilitation. The application of gravity compensation would not be interfering with the temporal characteristics of muscle activation, which are often impaired in stroke patients (Chae et al., 2002). In this way, patients would be able to repeat more movements and/or attend longer or more frequent sessions (i.e., higher training intensity) than in a situation without gravity compensation, which is beneficial in post-stroke rehabilitation (Kwakkel et al., 1999). Future studies should further examine this potential of gravity compensation for stroke rehabilitation.

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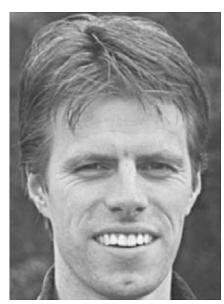

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