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Task variation during simulated, repetitive, low-intensity work – influence on manifestation of shoulder muscle fatigue, perceived discomfort and upper-body postures

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Work-related musculoskeletal disorders are increasing due to industrialisation of work processes. Task variation has been suggested as potential intervention. The objectives of this study were to investigate, first, the influence of task variation on electromyographic (EMG) manifestations of shoulder muscle fatigue and discomfort; second, noticeable postural shoulder changes over time; third, if the association between task variation and EMG might be biased by postural changes. Outcome parameters were recorded using multichannel EMG, Optotrak and the Borg scale. Fourteen participants performed a one-hour repetitive Pegboard task in one continuous and two interrupted conditions with rest and a pick-and-place task, respectively. Manifestations of shoulder muscle fatigue and discomfort feelings were observed throughout the conditions but these were not significantly influenced by task variation. After correction for joint angles, the relation between task variation and EMG was significantly biased but significant effects of task variation remained absent.

Practitioner Summary: Comparing a one-hour continuous, repetitive Pegboard task with two interrupted conditions revealed no significant influences of task variation. We did observe that the relation between task variation and EMG was biased by posture and therefore advise taking account for posture when investigating manifestations of muscle fatigue in assembly tasks.

Keywords: task variation; multichannel electromyography; posture; shoulder; muscle fatigue

1. Introduction

Work-related musculoskeletal disorders are a major problem in industrialised countries; disorders in the shoulder region comprise a large part of this problem (Buckle and Devereux 2002). These shoulder disorders are caused by multiple work-related risk factors such as repetitiveness and monotony of work processes (Hagberg et al. 1995; Buckle and Devereux 2002). To date, the exposure to these risk factors has increased due to automation, standardisation and rationalisation (Mathiassen 2006). The detrimental effect of these risk factors may be attenuated by implementing task variation (e.g. Mathiassen 2006).

Various types of task variation such as temporal variation, activity variation and additional breaks have been shown to reduce and prevent shoulder symptoms in repetitive, monotonous work (Henning et al. 1997); however, a recent review indicated that the effects of temporal and activity variation on, in particular, electromyography (EMG) and perceived discomfort are not very clear (Luger et al. 2014). The three papers on temporal variation revealed no positive effects on EMG and perceived discomfort; 14 studies on activity variation did not control for the amount and intensity of the work, which made it impossible to draw conclusions on the effect of activity variation alone. This means that the current practice or policy of introducing variation lacks empirical support and relies solely on theories like the so-called Cinderella hypothesis (Hägg 1991). Clearly, there still is a need for better understanding about whether and when (and when not) task variation may help preventing the manifestation of muscle fatigue, as it is considered an important precursor of shoulder disorders and musculoskeletal disorders at large.

Several methodological issues might explain the ambiguous EMG results in studies on task variation. First, bipolar EMG was used to determine the manifestation of muscle fatigue. Multichannel EMG, by contrast, allows for measuring larger muscle surfaces and, by this, is expected to provide more insight into the functionally different muscle compartments within the same muscle and into activation centres (Holtermann, Roeleveld, and Karlsson 2005; Staudenmann et al. 2010). Second, conventional EMG parameters including amplitude and frequency may not be the most suitable parameters. For instance, muscle fatigue would lead to heterogeneous muscle activity, which can be represented by an increase in

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sample entropy, a decrease in amplitude variability and a shift in the spatial centre of activity. Changes in these parameters may indicate the ability to use different motor strategies, reflected as motor flexibility (Jensen and Westgaard 1995; Palmerud et al. 1995; Richman and Moorman 2000; Madeleine and Madsen 2009; Madeleine 2010; Srinivasan and Mathiassen 2012). Third, postural changes may be reflected in conventional EMG parameters and mask subtle changes in the EMG manifestations of muscle fatigue (Andersson et al. 1974; Côté et al. 2005). In fact, this close relation between posture and muscle activity challenges the interpretation of EMG in general. To our best knowledge, no task variation studies have incorporated multichannel EMG or variability measures together with postural measures.

The purpose of this study was to investigate the effects of task variation on the manifestation of shoulder muscle fatigue, local perceived discomfort and 3D kinematics of upper body segments during a one-hour realistic, repetitive assembly task. Our specific research questions were:

(1) What is the effect of task variation on conventional EMG parameters of muscle fatigue and local perceived discomfort over time?
(2) What is the effect of task variation on more sophisticated EMG parameters like sample entropy, amplitude variability and centre of activity over time?
(3) Are there noticeable changes in posture over time? What is the effect of task variation on EMG parameters when controlled for postural changes?

We expect that at low intensity the more dynamic character of the active task variation in itself could influence fatigue development. Therefore, we have chosen to keep the weight of the box as low as possible, and used a crate for this purpose. We defined manifestation of muscle fatigue as a combined increase in EMG amplitude and decrease in frequency. We expect to find fatigue developments in all EMG parameters and perceived discomfort during the continuous condition but smaller fatigue developments in conditions with task variation because interruptions provide some rest for muscles or muscle compartments to recover. Regarding the posture of the subjects, we expect to see changes over time particularly for the neck flexion angle and the clavicle and scapula angles. Since it is known that posture and EMG are related, we furthermore expect these joint angles to slightly bias the effect of task variation on EMG.

2. Methods

2.1. Participants

Fourteen healthy, right-handed subjects (seven male and seven female, mean age 23.8 [SD 2.8] years, weight 69.2 [SD 10.2] kg, height 176.8 [SD 10.5] cm) volunteered to participate. None of the participants reported any history of musculoskeletal symptoms in the upper extremity for the previous six months. Subjects were not allowed to perform any strenuous exercise 24 h prior to the test period. Participants signed an informed consent after having been explained the aims and procedures of the experiment. The Ethics Committee of the Faculty of Human Movement Sciences approved the study.

2.2. Procedure

Subjects performed three one-hour conditions on one day, with 30-min rest periods in between. The order of the three conditions was systematically varied across subjects meaning that all possible combinations were assigned to both genders at least once. To familiarise the subjects with the task, a training session of about 5 min was performed before the start of the experiment. The training session ended when participants were able to set the correct pace to perform the task. Electromyographic reference contractions (ERC) were performed before the start of the first condition. Overall participation lasted for about 6 h.

The design of the experimental task was based on previous laboratory studies (Sundelin 1993; Henning et al. 1997; Sjörs et al. 2009; Sjøgaard et al. 2010), which simulated low-intensity assembly work using a Pegboard (Purdue Pegboard Model 32020; Lafayette Instrument Company, Lafayette, IN, USA; Figure 1). We applied two types of task variation in a work-rest ratio of 55:5 min. We have included a passive 1-min rest break and an active 1-min box-replacement task. The chosen work-rest ratio is based on previous research, as Dababneh, Swanson, and Shell (2001) found that frequent short interruptions are beneficial to workers wellbeing.

The subjects bimanually filled and emptied 12 holes (6 × 2 matrix) in a Pegboard, resulting in 24 repetitive movements in 30 s (the total cycle time). When a pin fell down, the participant was instructed to proceed with the Pegboard task without picking up the pin. Using the Pegboard task enabled us to standardise work cycles. Before the experiment started, sitting height was individually adjusted to a knee angle of 90° and working height was individually adjusted to an elbow flexion angle of slightly less than 90° and an upper arm elevation angle of about 30° relative to the vertical.

In all three conditions, subjects were instructed to perform the task for one-hour without leaning on the table. In the continuous condition the 30-s cycles were continuously repeated; in total, 120 cycles were performed in one-hour.

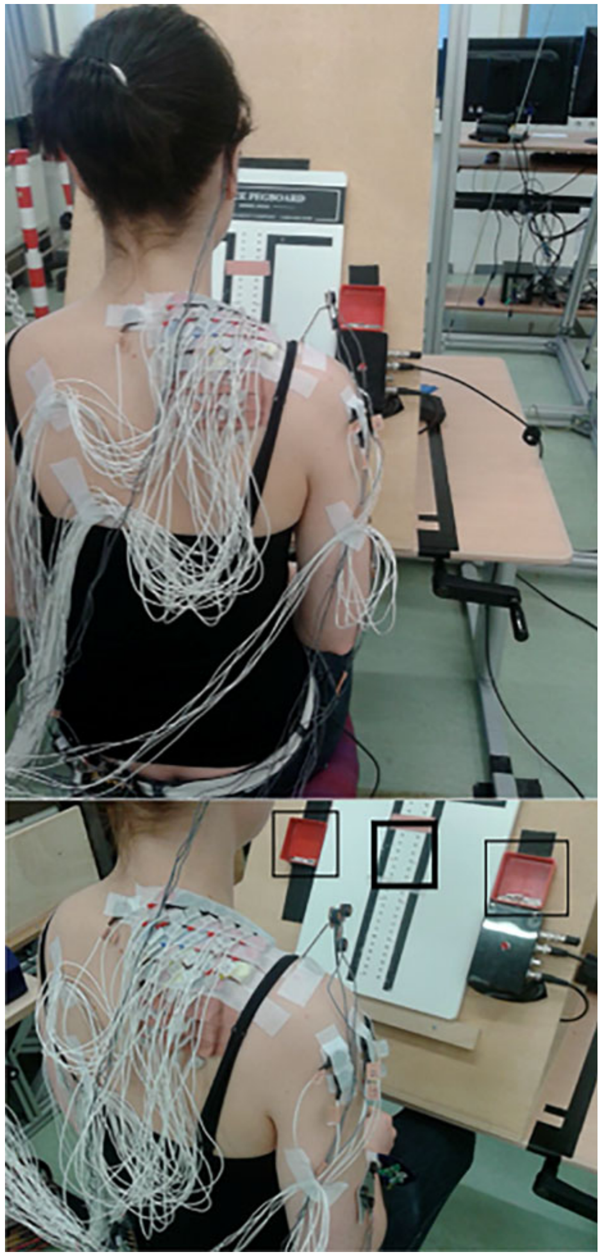

Figure 1. A sample subject performing the Pegboard task. Holes were bimanually filled and emptied with pins (1.3 g) from and to the two bins next to the Pegboard (see the two thin rectangles). A restricted part of the Pegboard was used (6 × 2 matrix, see the thick rectangle) to keep the task as static as possible. At the right shoulder (M. Trapezius), muscle activity is measured using two grids of electrodes.

In the two interrupting conditions, 22 cycles (11 min) were followed by a one-minute rest break (I_{break}) keeping the hands on the lap or followed by a one-minute dynamic work task (I_{work}) by picking and placing an empty crate (1.9 kg, 40 × 30 × 30 cm [l × b × h]) five times from the left-side to the right-side while seated. This procedure, 22 cycles followed by a one-minute interruption, was repeated five times.

2.3. Measurements

2.3.1. Electromyographic reference contractions

Before the start of the experiment, ERC measurements were performed for 5 s. The subject was instructed to stand upright with the arm vertical, the elbow flexed forward in 90° and the forearm in a neutral position (i.e. the hands vertical with the thumbs pointing upward). The subject was instructed to exert maximum force with the right arm in the upward direction, while holding a handle. During three ERC attempts, muscle activity was recorded for the five muscle subparts. The one-second time period with the highest mean activity level determined the ERC.

2.3.2. *Electromyography*

EMG signals from five muscle subparts at the right dominant side were recorded, including the M. Trapezius Transversus (MTT), M. Trapezius Descendens (MTD), M. Deltoideus Clavicularis (MDC), M. Deltoideus Acromialis (MDA) and M. Deltoideus Spinalis (MDS) using monopolar Ag/AgCl surface electrodes (KendallTM H69P Cloth Electrodes, Covidien, Zaltbommel, The Netherlands). We have chosen the Trapezius subparts because these are considered to be one of the most affected muscles in the shoulder region (Westgaard, Jansen, and Jensen 1996). The Trapezius muscle is involved in stabilising the shoulder and the primary mover in shoulder elevation (Inman, Saunders, and Abbott 1944). We have chosen the Deltoid subparts because these are constantly used during humerus flexion and elevation as part of our Pegboard task. The midpoints of the five muscle subparts were determined as halfway the line connecting the origin of the muscle with their insertion, as recommended by Hermens et al. (2000). Electrode positions were marked on the skin with a pencil. Before electrode placement, the skin was shaved, scrubbed with sandpaper and cleaned with alcohol. On the MTT and MTD, the electrodes were placed in a 2D array in line with the fibre direction with an inter electrode distance (IED) of 15 mm (Figures 2 and 3). On the MDC, MDA and MDS, two monopolar Ag/AgCl surface electrodes were placed in a bipolar configuration on both sides of the midpoints in line with the fibre direction with an IED of 20 mm. A wet wristband on the left Ulnar Styloid served as ground electrode.

EMG signals were online analogue high-pass filtered (10 Hz), amplified with a 128-channel amplifier (REFA, TMS International B.V., Enschede, the Netherlands) and sampled at 2048 Hz. EMG signals were recorded every other cycle for 30 s. Using Matlab (The Mathworks Inc., Natwick, MA, USA), the bipolar derivations of the recordings were computed and filtered offline with a bidirectional second-order Butterworth filter (10–400 Hz).

For all bipolar derivations, the mean amplitude over each cycle was determined by the average rectified value (ARV) and normalised to 100% ERC. The median frequency (MdPF) over each cycle was determined via the power spectral density of the raw EMG that was estimated using Welch's periodogram method (Welch 1967). We calculated the mean ARV and MdPF for each muscle using all bipolar derivations of the specific muscle. This means that we used 16 and 24 bipolar derivations to calculate the mean ARV and MdPF for the M. Trapezius Transverus and M. Trapezius Descendens, respectively. For the subdivisions of both Trapezius muscle parts, we used the bipolar derivations that were located in the specific subdivision (see Figures 2 and 3) to calculate their mean ARV and MdPF. For each condition and subject, we estimated the linear slope of the ARV and MdPF over time for the whole muscle and for the subdivisions of the MTT and MTD.

Topographical mapping allowed us to determine the spatial centre of activity (CoA) in the MTT and MTD in the x- and y-direction (CoA-x and CoA-y). We determined the position of the CoA and expressed this position in millimetres (Falla and Farina 2007). The regularity of the EMG signals was determined by the sample entropy (SampEn) for all five muscle subparts (Lake et al. 2002). The variability of the ARV in the five muscle subparts in each measurement sample was

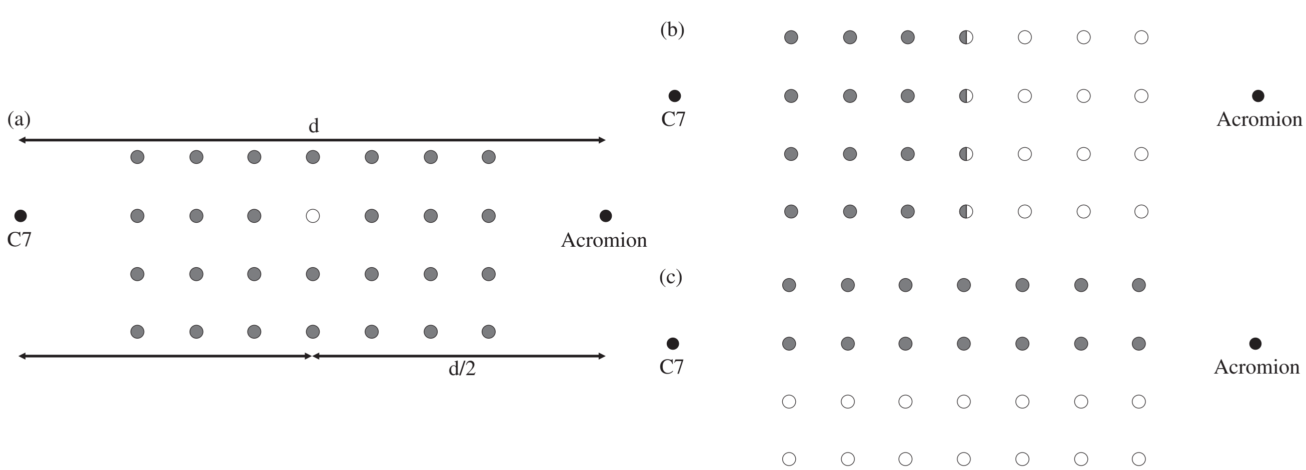

Figure 2. Schematic electrode array configuration (4 × 7) of the M. Trapezius Descendens with an interelectrode distance (IED) of 15 mm. (a) The midpoint of the array is defined as half of the distance (d/2) between the Cervical Vertebra C7 and the Acromion, depicted as a white circle; (b) Medial and Lateral subdivisions of the grid depicted in grey and white, respectively; (c) Cranial and caudal subdivisions of the electrode grid depicted in grey and white, respectively.

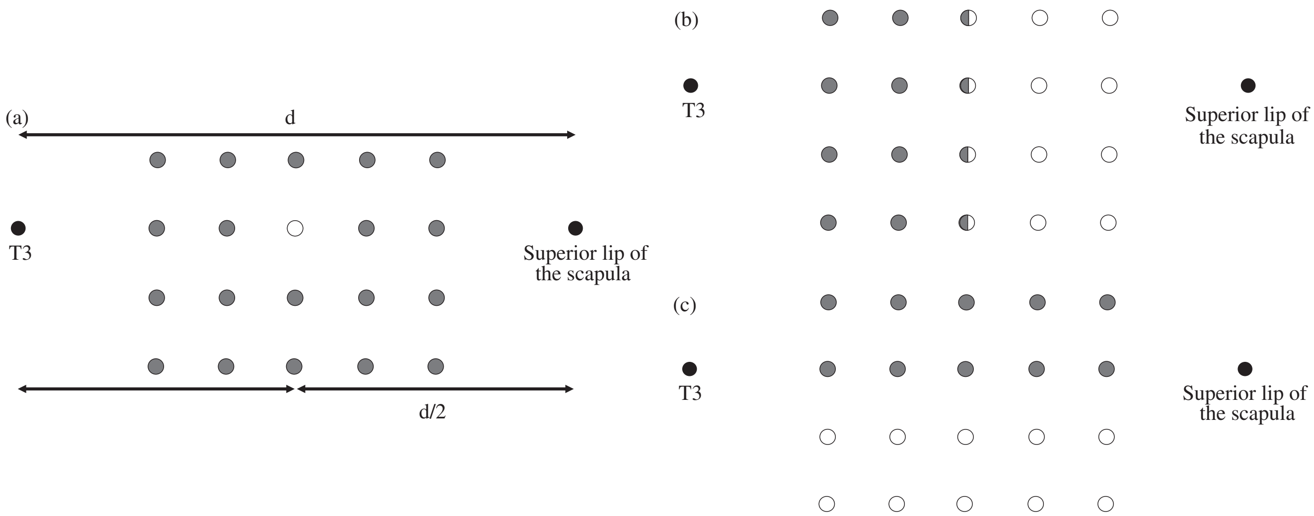

Figure 3. Schematic electrode array configuration (4 × 5) of the M. Trapezius Transversus with an IED of 15 mm. (a) The midpoint of the array is defined as half of the distance (d/2) between the Thoracic Vertebra T3 and the Superior Lip of the Scapula, depicted as a white circle; (b) Medial and Lateral subdivisions of the grid depicted in grey and white, respectively; (c) Cranial and caudal subdivisions of the electrode grid depicted in grey and white, respectively.

determined by its standard deviation (SD). The linear slopes for each condition of the sample entropy (SampEn), amplitude variability (ARV-SD) and centre of activity coordinates (CoA-*x*, CoA-*y*) were calculated using the sixty 30-s samples to indicate their change over time.

2.3.3. *Rating of perceived discomfort*

Subjects were asked to rate the perceived discomfort (RPD) in their right shoulder area on the CR-10 Borg scale, which ranged from 0 (no discomfort) to 10 (extremely strong, almost maximal; Borg 1982). RPD of the right shoulder was assessed at the start of and every 5 min during each condition. The linear slope of the RPD served to indicate its change over time.

2.3.4. *Kinematics*

Postures of the right upper body were recorded using Optotrak (Northern Digital Inc, Waterloo, Ontario, Canada). Cluster markers were placed on upper body segments including the head, trunk, right scapula, right upper arm and right forearm. Anatomical landmarks were marked with a pencil and visually probed before the start of the experiment (see Wu et al. 2005 for an overview). To determine the glenohumeral rotation centre (GH) of the upper arm, the subject performed a circular movement with the right arm before the start of the experiment. We estimated GH using an instantaneous helical axis algorithm (Veeger et al. 1997).

Recording of the cluster markers during the experimental conditions lasted for 30 s and was repeated every other cycle. The exact same time intervals were also used for EMG recordings. Recordings were sampled at 100 Hz by two camera bars placed in front of and diagonally right behind the subject. With Matlab, relative joint angles were calculated using the ISB recommendations (Wu et al. 2005): trunk flexion-extension, scapula re-protraction, scapula anterior–posterior tilt, scapula medial–lateral rotation (Veeger et al. 2003), clavicle re-protraction, clavicle elevation–depression, humerus forward flexion, humerus elevation, elbow flexion and neck flexion–extension. We calculated the linear slopes of these joint angles to determine their change over time. A similar procedure was applied on our EMG data. We related each joint angle with specific target muscles: (1) MTT with clavicle re-protraction, scapula re-protraction and scapula anterior–posterior tilt; (2) MTD with clavicle elevation–depression, scapula lateral rotation, humerus elevation and neck flexion; (3) the Deltoideus subparts with humerus flexion and elevation.

2.4. Statistical analysis

Inspection of the data indicated that all variables were normally distributed except for the ARV-SD. The slopes of all variables were tested if they differed from zero using one sample *t*-tests with bootstrapping to investigate a fatigue effect

(Field 2013). We used generalised estimating Equation (GEE) models with an exchangeable correlation matrix to allow for assessing repeated measures within participants. Using the GEE models, we tested whether there was an effect of task variation (predictor) on the slopes of ARV, MdPF, RPD, SampEn, ARV-SD and CoA (outcomes), displayed in Equation (1). The values B_1 and B_2 indicate the difference in slope between I_{break} and the continuous condition (B_1) and between I_{work} and the continuous condition (B_2); the p-values indicate whether these differences are significantly different from zero.

$$\text{Outcome} = B_0 + B_1 * (I_{\text{break}}\ \text{vs. Continuous}) + B_2 * (I_{\text{work}}\ \text{vs. Continuous}) \tag{1}$$

$$\text{Outcome} = B_0 + B_1 * (I_{\text{break}}\ \text{vs. Continuous}) + B_2 * (I_{\text{work}}\ \text{vs. Continuous}) + B_3 * \text{Covariate} \tag{2}$$

To test whether certain joint angles biased the relation between task variation and EMG, we extended the GEE models with the joint angle's slope as a covariate (Equation (2)). Associations are biased by posture if the values B_1 and B_2 from Equation (1) change at least 10% when the covariate is added to the model, as in Equation (2) (Skelly, Dettori, and Brodt 2012). The value B_3 indicates how much the slope (outcome) changes per degree of joint angle change, corrected for the effects of condition (B_1 and B_2). We used B_3 as a covariate and did not interpret its specific values. Significance was accepted at $p < 0.05$. All statistical analyses were performed with SPSS (IBM SPSS Statistics 21.0).

3. Results

3.1. Conventional fatigue parameters

The one sample t-tests revealed significant increases of ARV in all conditions for the Trapezius subparts and significant decreases of MdPF in most conditions for all five muscle subparts over time (Figure 4) as well as for the muscle compartments of both Trapezius subparts (Figure 5). These significant changes indicate the development of muscle fatigue in the MTT and MTD in some conditions. The GEE models revealed a significant effect of task variation (see Appendix A): in I_{break} the MdPF decreased more than in the continuous condition ($p < 0.05$) for the MTD, which, however, was not significant in its muscle compartments.

In all three conditions, RPD increased significantly over time reaching mean levels of 5.4 ($\pm$2.0) in the continuous condition, 4.8 ($\pm$2.0) in I_{break} and 4.9 ($\pm$2.2) in I_{work} at the end of the conditions. While I_{break} did not differ significantly from the continuous condition, there was a significantly steeper slope for the continuous condition than for I_{work} ($p < 0.05$, see Appendix A).

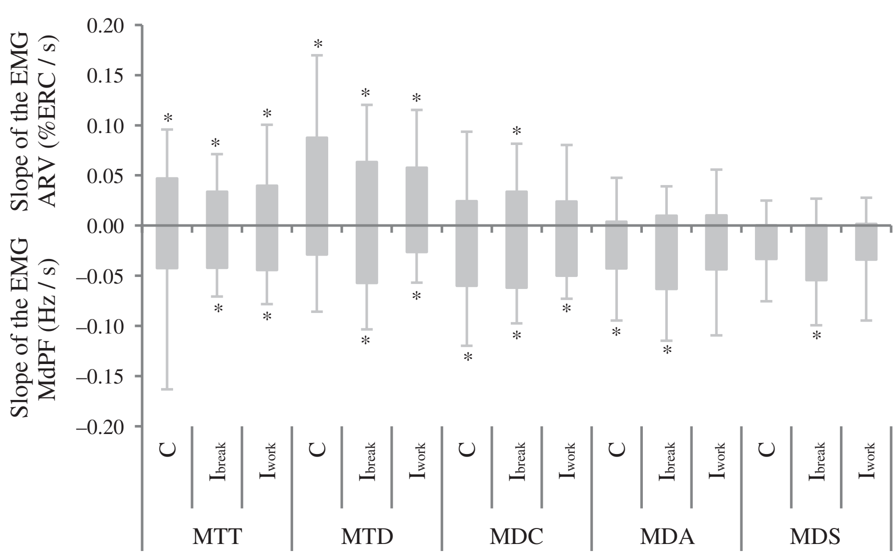

Figure 4. Slopes of conventional EMG parameters. Positive bars reflect the EMG amplitude (ARV) in %ERC/s; negative bars represent the EMG median frequency (MdPF) in Hz/s. Bars represent the slope for the M. Trapezius Transversus (MTT), M. Trapezius Descendens (MTD), M. Deltoideus Clavicularis (MDC), M. Deltoideus Acromialis (MDA) and M. Deltoideus Spinalis (MDS) during the continuous condition (C), I_{break} and I_{work}. Error bars represent the SD between subjects. Bars marked with an asterisk (*) are significantly different from zero ($p < 0.05$).

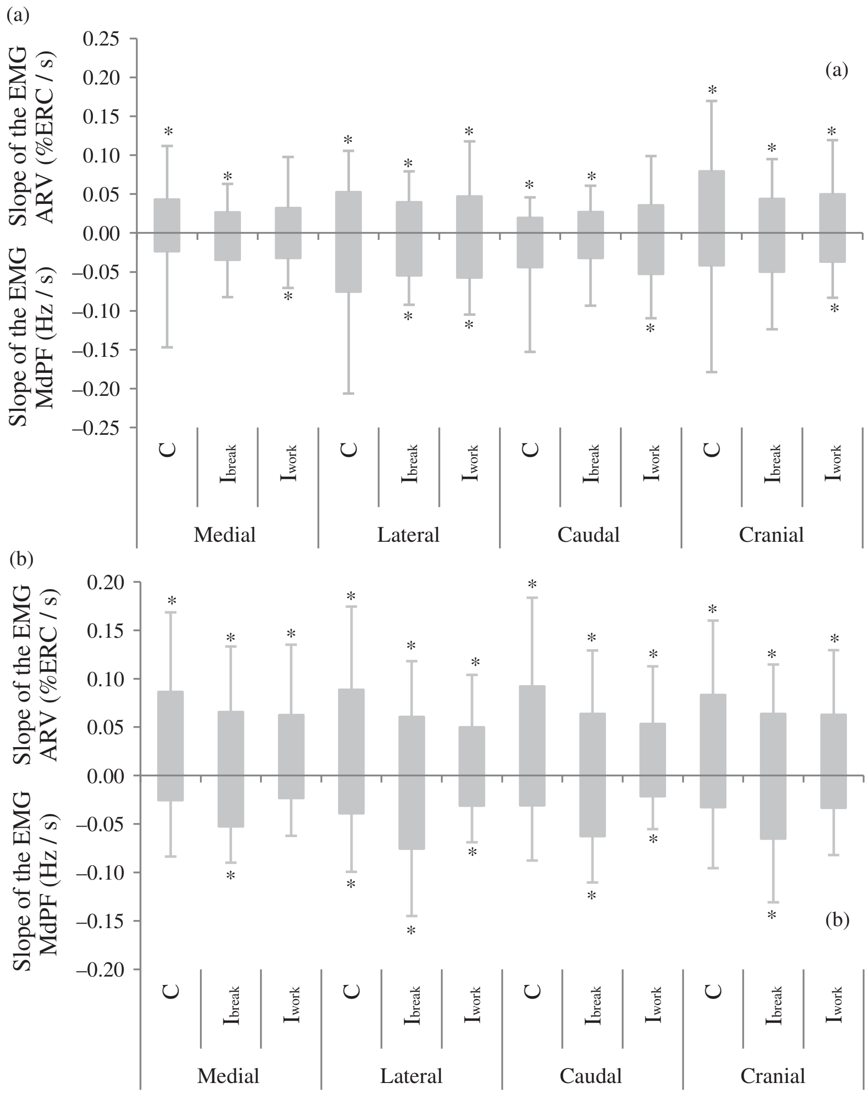

Figure 5. Slopes of conventional EMG parameters in muscle compartment of the Trapezius subparts. (a) compartments of the M. Trapezius Transversus; (b) compartments of the M. Trapezius Descendens. Positive bars represent the slope for the EMG ARV in % ERC/s and negative bars represent the slope for the EMG MdPF in Hz/s for the continuous condition (C), I_{break} and I_{work}. Error bars represent the SD between subjects. Bars marked with an asterisk (*) are significantly different from zero ($p < 0.05$).

3.2. Sophisticated EMG manifestations of muscle fatigue

We found a significant decrease in SampEn in some conditions of the MDC and MDA over time (Figure 6a), but there were no differences between conditions (see Appendix A). ARV-SD, by contrast, increased significantly in all conditions of the MTD (Figure 6b) as well as in its compartments (Figure 7b).The GEE models revealed significant effects of task variation for muscle compartments of the Trapezius Descendens: ARV-SD of the lateral part was significantly lower in I_{break} than in the continuous condition and ARV-SD of the caudal part was significantly lower in I_{work} than in the continuous condition (see Appendix A). The CoA revealed no significant shifts over time and no effects of task variation (see Appendix A).

3.3. Postural changes

The joint angles which showed strongest changes over time included clavicle retraction, clavicle elevation, humerus flexion and neck flexion: clavicle retraction significantly increased in the continuous condition and I_{break}; the clavicle elevation

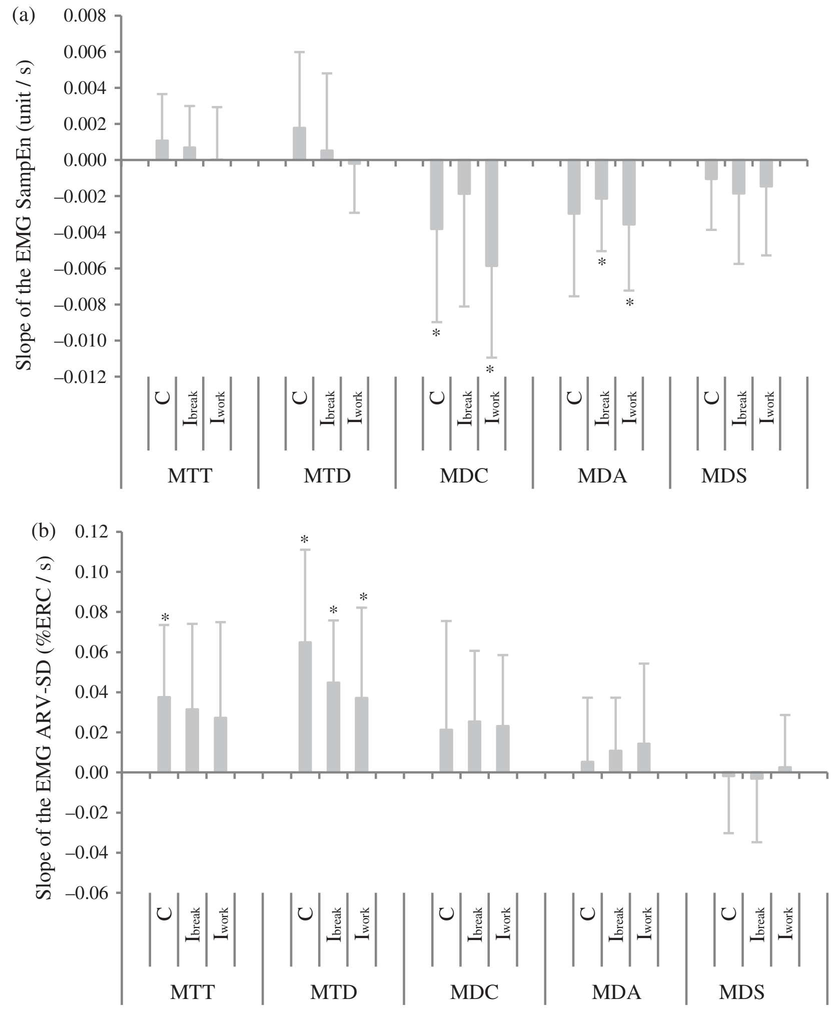

Figure 6. Slopes of sophisticated EMG parameters. (a) EMG sample entropy (SampEn) in unit/s; (b) EMG amplitude variability (ARV-SD) in %ERC/s. Bars represent the slope for the M. Trapezius Transversus (MTT), M. Trapezius Descendens (MTD), M. DeltoideusClavicularis (MDC), M. DeltoideusAcromialis (MDA) and M. DeltoideusSpinalis (MDS) during the continuous condition (C), I_{break} and I_{work}. Error bars represent the SD between subjects. Bars marked with an asterisk (*) are significantly different from zero ($p < 0.05$).

angle increased significantly in the continuous condition; humerus flexion decreased significantly in the continuous condition and neck flexion changed from slight extension towards flexion in all conditions (Figure 8).

With the extended GEE models (Equation (2)) we tested whether certain postures biased the association between task variation and EMG outcome parameters (see Appendix B). Associations were biased by posture if the coefficient of I_{break} versus the continuous condition (B_1) or I_{work} versus the continuous condition (B_2) changed at least 10% compared to the simpler model without covariates (Equation (1); Skelly, Dettori, and Brodt 2012). The effect of task variation on the CoA was hardly biased by posture (see Appendix B). The effect of task variation on MTT ARV, MdPF and ARV-SD was disturbed by clavicle retraction, scapula protraction and scapula anterior tilt; however, the effects of task variation on MTT EMG remained not significant after postural correction. For the MTD, the effect of task variation on ARV, MdPF, ARV-SD and SampEn was disturbed by clavicle elevation, scapula lateral rotation and humerus elevation. Correction for these postures also did not result in significant effects of task variation on MTD EMG. The associations between task variation and EMG outcomes in the three Deltoideus subparts were disturbed by both humerus flexion and elevation (see Appendix B). Humerus elevation clearly changed the coefficients B_1 and B_2 in all three Deltoideus subparts. Unadjusted, there were no significant effects of task variation on EMG; however, when corrected for humerus elevation, the ARV-SD in the Deltoideus Acromialis became significantly higher in I_{break} compared to the continuous condition ($p < 0.05$).

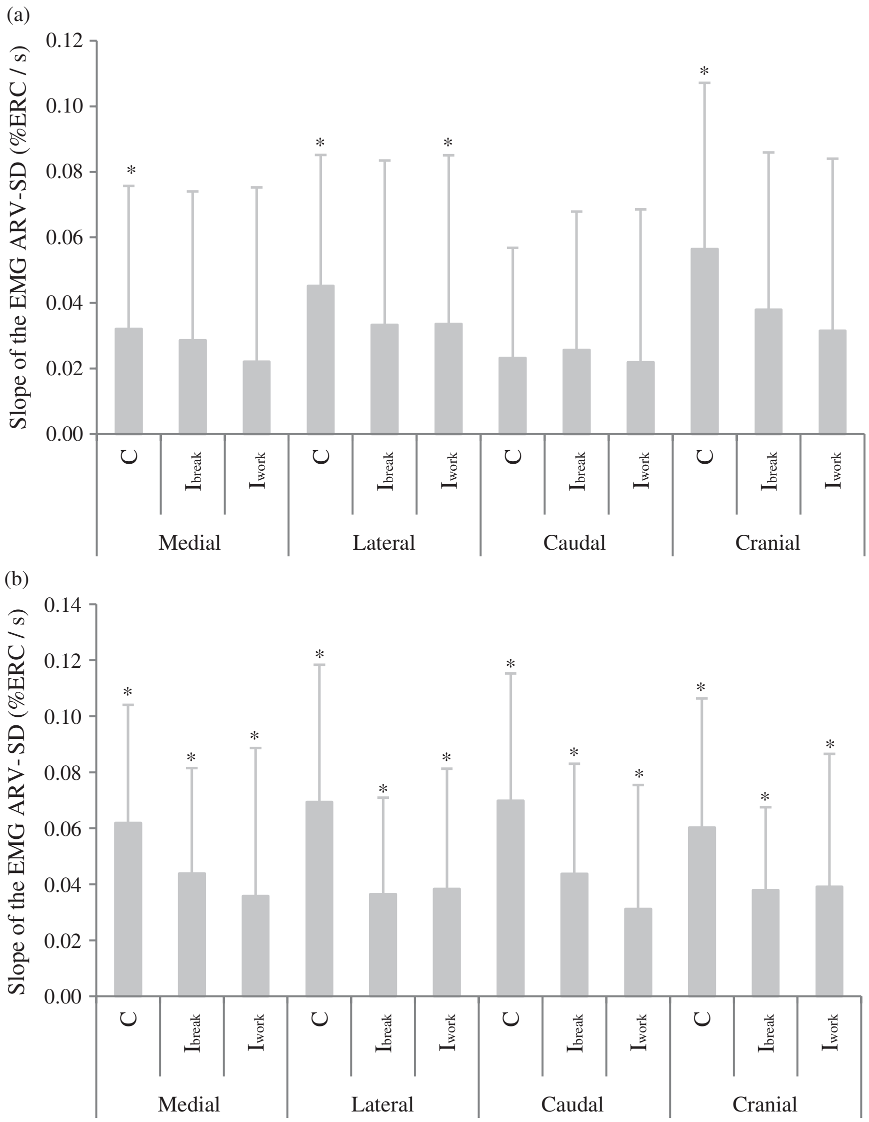

Figure 7. Slopes of the ARV-SD of muscle compartments of Trapezius subparts. (a) compartments of the M. Trapezius Transversus; (b) compartments of the M. Trapezius Descendens. Bars represent the slope for the EMG ARV-SD in %ERC/s for the continuous condition (C), I_{break} and I_{work}. Error bars represent the SD between subjects. Bars marked with an asterisk (*) are significantly different from zero ($p < 0.05$).

4. Discussion

With the present study we first aimed for investigating the effects of task variation on the conventional EMG manifestation of shoulder muscle fatigue and local perceived discomfort during a one-hour simulated, repetitive assembly task; however, task variation had no significant effect on these parameters. We also aimed for determining the effects of task variation on the manifestation of shoulder muscle fatigue reflected in EMG variability parameters, which, likewise, showed no significant results. Finally, we wanted to determine the upper body postures and their reflection in EMG parameters. We found that several joint angles biased the relation between task variation and EMG outcome parameters in different muscle subparts. Despite correcting for angles, effects of task variation remained not significant except for the ARV-SD in the M. Deltoideus Acromialis.

4.1. Task variation and conventional fatigue parameters

Our results showed that, with or without correction for posture, task variation did not influence the conventional manifestation of shoulder muscle fatigue, defined as a combined increase in EMG amplitude and decrease in median

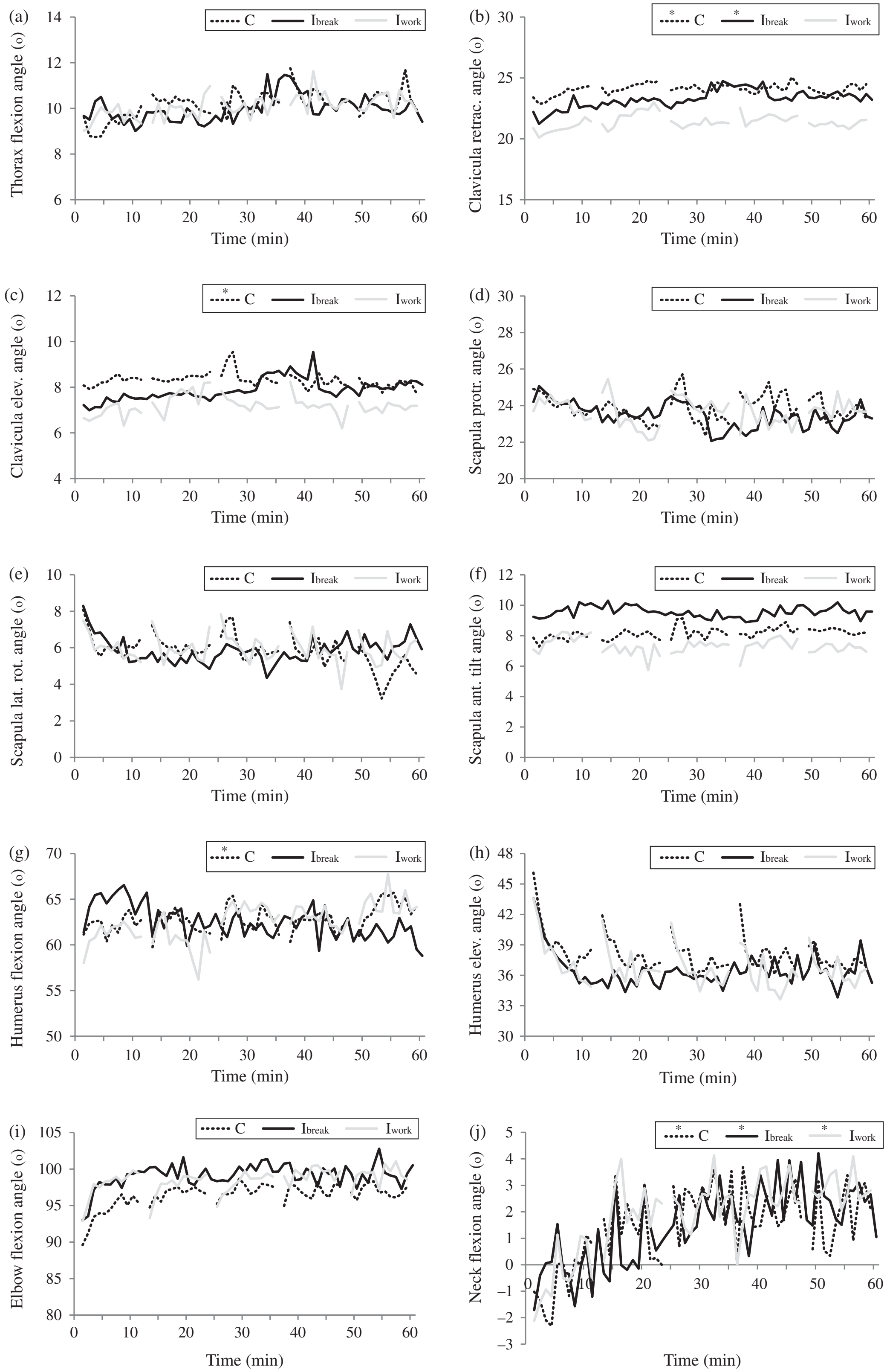

Figure 8. Change in joint angles over time during the continuous condition (C), I_{break} and I_{work}. (a) Thorax flexion; (b) Clavicle retraction; (c) Clavicle elevation; (d) Scapula protraction; (e) Scapula lateral rotation; (f) Scapula anterior tilt; (g) Humerus flexion; (h) Humerus elevation; (i) Elbow flexion angle; (j) Neck flexion. Lines marked with an asterisk (*) show a significant change over time ($p < 0.05$).

frequency (Basmajian and De Luca 1985). Supporting these results, several previous studies demonstrated that task variation does not influence the EMG amplitude and frequency (Byström, Mathiassen, and Fransson-Hall 1991; Rissén et al. 2002; Yassierli and Nussbaum 2007). It might be that task variation was too short to reduce the amount of active motor units (Moritani and Muro 1987). Iridiastadi and Nussbaum (2006) suggested that different muscle compartments or adjacent synergistic muscles are activated over time, which may have led to the absence of changes indicating muscle fatigue in the whole target muscle.

Local perceived discomfort decreased in response to task variation I_{work}. Several studies performed short interventions during a comparable fatigue protocol and showed that indicators of perceived fatigue improved after introducing passive or active interruptions (Henning et al. 1997; Galinsky et al. 2000; Dababneh, Swanson, and Shell 2001; Balci and Aghazadeh 2003; Yassierli and Nussbaum 2007; Yung, Mathiassen, and Wells 2012). Sundelin (1993) showed beneficial effects of short interruptions on both perceived fatigue and EMG manifestations of muscle fatigue. However, there are also studies, partly in line with our results, that showed no significant responses to passive or active interruptions (Mathiassen 1993; Mathiassen and Winkel 1996; Rissén et al. 2002; Iridiastadi and Nussbaum 2006). In line with the findings of Crenshaw, Djupsjöbacka, and Svedmark (2006), we found that active task variation provides more opportunities for the musculoskeletal system to recover, represented as significantly less discomfort development.

4.2. Task variation and sophisticated EMG manifestations of muscle fatigue

The study results showed that, both with and without correction for posture, task variation did not influence the more sophisticated EMG parameters. We quantified the sophisticated manifestation of muscle fatigue in different ways: first, an increase in SampEn; second, a decrease in ARV-SD; third, a shift of the CoA. The EMG's SampEn decreased more in the task variation conditions than in the continuous condition although not significantly. This difference between conditions, which occurred in all five muscle subparts, may indicate that task variation induces higher regularity of muscle activation. This is in contrast with the literature, which indicates that muscle fatigue would result in irregular muscle activity, reflecting the ability to use different motor strategies (Costa, Goldberger, and Peng 2002; Möller et al. 2004; Madeleine and Madsen 2009). The ARV-SD reflects the amount of variability. This parameter increased significantly in several conditions and muscle subparts over time. The increased ARV-SD over time may be either a result of fatigue or a compensatory strategy to prevent fatigue (Srinivasan and Mathiassen 2012). Increased ARV-SD may protect employees from developing fatigue (Madeleine, Voigt, and Mathiassen 2008) and it may be advantageous for the Cinderella motor units which will not be constantly and equally activated to prevent overuse and fatigue (Hägg 1991). ARV-SD was not affected by task variation, which is in contrast with the findings of Samani et al. (2009) indicating more amplitude variability in work with short interruptions. With respect to the shift in the CoA, we found no significant effects of fatigue or task variation. This is not in line with the results of Samani et al. (2010), who found that the CoA significantly shifted in a condition with active pauses compared to a condition without pauses. Their experimental conditions – no breaks, passive breaks and active breaks – are quite similar to our conditions except for the computer task and shorter 10-min duration. We indeed observed another location of the CoA in the MTT: the CoA for I_{work} was located more lateral and cranial compared to the continuous condition and I_{break}, which, although not significant, may reflect an increased load applied to the upper part of the MTT. This shift may also indicate functional subdivisions in the Trapezius subparts during low level activation (Mathiassen, Winkel, and Hägg 1995; Jensen and Westgaard 1997), which becomes apparent after peripheral and central fatigue mechanisms as increasing the amount of variability may be a strategy to cope with fatigue (Farina et al. 2008).

4.3. Postural changes and EMG parameters

The present findings showed that clavicle retraction, clavicle elevation and neck flexion significantly increased and humerus flexion significantly decreased over time (Figure 8). It remains unclear, however, whether the changes in joint angles indicate a compensation strategy to prevent (further) fatigue or are the result of muscle fatigue. Several studies indicate a causal relation that the shoulder girdle's position and movement are affected by any fatigued shoulder muscle (Mclean 2005; Ebaugh, Mcclure, and Karduna 2006) especially by the current task (Fuller et al. 2009).

The two Trapezius subparts are especially involved in stabilising the scapula to improve functional efficiency. The MTT also retracts the scapula and clavicle, which makes it not surprising that clavicle retraction, scapula protraction and scapula anterior tilt biased the effect of task variation on EMG ARV, MdPF and ARV-SD (see Appendix B). Differences between the continuous condition and I_{break} still remained very small and not significant. On the contrary, when corrected for posture, I_{work} tended to reduce amplitudes and increase frequencies which may imply less fatigue development when repetitive work is interrupted by another work activity. The MTD, on the other hand, acts as an elevator of the scapula and clavicle, which is reflected in the disturbing effect of clavicle elevation and scapula lateral rotation on the relation between task variation and EMG ARV, MdPF, ARV-SD and SampEn (see Appendix B). Both joint angles decreased the differences between the continuous condition and I_{break}, where I_{work} showed a smaller tendency towards muscle fatigue because of a higher frequency compared to the continuous condition. Since the medial part of the MTD is attached to the nuchal ligament and superior nuchal line at the base of the skull, we would expect that neck flexion biases the effect of task variation on EMG parameters. Although neck flexion significantly changed from slight extension towards flexion over time, a relatively small change ($\pm 3°$), it did not bias the effect of task variation on EMG. From an anatomical point of view, we expected a bias of humerus elevation on the association between task variation and MTD EMG parameters. We found this bias, indeed, and it resulted in smaller effects of task variation for ARV and ARV-SD (see Appendix B). This observation may indicate that humerus elevation had only minor influence because effects of task variation are still not significant after correction. In line with this tendency, Arvidsson et al. (2012) and Bagg and Forrest (1986) observed an increased MTD activity along with an increased upper arm elevation angle.

Regarding the Deltoideus subparts, humerus elevation is mainly associated with MDA activity and flexion with MDC activity. Adding humerus elevation as a covariate resulted in a higher ARV-SD in the interrupted conditions. Particularly focusing on the MDA, I_{work} tended to differ most with the continuous condition because of an increased ARV and MdPF which is ambiguous with regard to the manifestation of muscle fatigue. The only significant change that occurred after correction for posture was a larger ARV-SD increase in I_{break} compared to the continuous condition ($p < 0.05$). With respect to humerus flexion, we expected it to mainly disturb the effect of task variation on MDC EMG; however, we found that this relation was influenced less than for MDA and MDS. The association between task variation and EMG in the Deltoideus subparts, when statistically corrected for both humerus angles, tended to an increased ARV-SD reflecting greater motor flexibility. This is considered healthier (Jensen and Westgaard 1995). Conventional EMG parameters, on the other hand, did not show such advantageous changes after correcting the relation between task variation and Deltoideus EMG for humerus flexion and elevation.

5. Limitations and implications

We aimed for more insight into the effects of task variation on manifestations of muscle fatigue by using multichannel EMG combined with 3D kinematics during a repetitive, simulated assembly task. This study is not without limitations. We simulated repetitive assembly work by performing a static pick-and-place task on a Pegboard, which does not really mimic a real-life working situation. One can imagine that in real-life people may perform similar movements but the requested accuracy level, time pressure and number of constraints will differ. Particularly, the combination between accuracy level and number of constraints certainly differs from realistic working conditions because precise tasks are usually less constrained and last longer than 30 s to avoid fatal production errors as much as possible. Furthermore, performing seated tasks on a stool may also not be representative for most of the workplaces. Also, this study interpreted the results of a group of young and healthy subjects, where a working population will be more diverse. Finally, the three one-hour conditions investigated were too short to extrapolate our findings directly to a full working day. These limitations alert caution in the extrapolation of the current results to real life work situations.

6. Conclusions and practical relevance

Our results could not prove that task variation, in the setting we provided, influences EMG manifestation of shoulder muscle fatigue or local perceived discomfort. Our expectation that task variation would have a positive effect on muscle fatigue has not been supported by the current results, which may imply that the task variation applied at this intensity level was negligible. There may be a power problem due to the number of subjects, increasing the chance on a type II error. Post hoc sample size calculations revealed that we would need at least 160 subjects for the EMG results to become significantly influenced by task variation. This number is unrealistically large considering the size of the effect and it remains questionable whether this effect is practically relevant. We wonder whether EMG is the correct method to study the effectiveness of interventions during low intensity work. From this study, and in line with earlier studies, it seems that the psychological parameters are much more sensitive for task variation. Therefore, we should consider adding more extensive psychological fatigue indicators, such as the Swedish Occupational Fatigue Index questionnaire (SOFI; Åhsberg 1997).

Nevertheless, we have seen that the effect of task variation on EMG outcome parameters was biased by several postural changes in the shoulder region, but this did not result in significant effects of task variation. Especially the clavicle elevation, scapular lateral rotation and humerus elevation angles were important covariates. This indicates that when investigating EMG manifestation of muscle fatigue during repetitive assembly tasks, we should take account for postural changes over time because these bias manifestations of muscle fatigue and potential interventions alike. Our results, however, do not provide evidence about a potential (causal) relation between posture and EMG manifestation of fatigue, which leaves room for further research.

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Disclosure statement

No potential conflict of interest was reported by the authors.

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Appendix A

Table A1. Statistical results of the GEE models for the effect of task variation on the slopes of the EMG amplitude (ARV) and median frequency (MdPF), centre of activity (CoA), amplitude variability (ARV-SD), sample entropy (SampEn), rating of perceived discomfort (RPD) for I_{break} or I_{work} versus the continuous condition (C).

		I_{break} (compared to C)		I_{work} (compared to C)	
Muscle	Outcome variable	B_1	95%-CI	B_2	95%-CI
MTT Whole	ARV	−0.013	[−0.036; 0.009]	−0.007	[−0.055; 0.040]
	MdPF	0.000	[−0.057; 0.058]	−0.002	[−0.059; 0.055]
	CoA-x	0.000	[−0.001; 0.001]	0.000	[−0.002; 0.002]
	CoA-y	0.001	[−0.000; 0.003]	0.001	[−0.001; 0.004]
	ARV-SD	−0.006	[−0.025; 0.013]	−0.010	[−0.046; 0.025]
	SampEn	0.000	[−0.002; 0.001]	−0.001	[−0.004; 0.001]
MTT Medial	ARV	−0.016	[−0.050; 0.017]	−0.011	[−0.070; 0.048]
	MdPF	−0.011	[−0.066; 0.044]	−0.009	[−0.074; 0.057]
	ARV-SD	−0.004	[−0.027; 0.020]	−0.010	[−0.046; 0.026]
MTT Lateral	ARV	−0.013	[−0.037; 0.011]	−0.005	[−0.053; 0.042]
	MdPF	0.021	[−0.050; 0.092]	0.018	[−0.042; 0.078]
	ARV-SD	−0.012	[−0.033; 0.010]	−0.012	[−0.050; 0.027]
MTT Caudal	ARV	0.008	[−0.012; 0.027]	0.016	[−0.024; 0.057]
	MdPF	0.012	[−0.067; 0.091]	−0.009	[−0.066; 0.049]
	ARV-SD	0.002	[−0.015; 0.020]	−0.001	[−0.031; 0.028]
MTT Cranial	ARV	−0.035	[−0.071; 0.000]	−0.030	[−0.098; 0.039]
	MdPF	−0.008	[−0.058; 0.041]	0.005	[−0.061; 0.071]
	ARV-SD	−0.018	[−0.043; 0.006]	−0.025	[−0.070; 0.021]
MTD Whole	ARV	−0.024	[−0.071; 0.022]	−0.030	[−0.089; 0.029]
	MdPF	−0.032	[−0.062; −0.001]*	0.005	[−0.016; 0.025]
	CoA-x	0.000	[−0.001; 0.001]	0.000	[−0.001; 0.001]
	CoA-y	0.000	[−0.001; 0.000]	−0.001	[−0.001; 0.000]
	ARV-SD	−0.024	[−0.051; 0.002]	−0.028	[−0.058; 0.001]
	SampEn	−0.001	[−0.003; 0.001]	−0.002	[−0.005; 0.002]
MTD Medial	ARV	−0.021	[−0.068; 0.026]	−0.024	[−0.086; 0.038]
	MdPF	−0.027	[−0.056; 0.002]	0.002	[−0.025; 0.030]
	ARV-SD	−0.018	[−0.047; 0.011]	−0.026	[−0.058; 0.006]
MTD Lateral	ARV	−0.028	[−0.079; 0.023]	−0.039	[−0.100; 0.022]
	MdPF	−0.036	[−0.078; 0.005]	0.008	[−0.023; 0.039]
	ARV-SD	−0.033	[−0.062; −0.004]*	−0.031	[−0.065; 0.003]
MTD Caudal	ARV	−0.028	[−0.076; 0.020]	−0.039	[−0.103; 0.026]
	MdPF	−0.032	[−0.068; 0.004]	0.010	[−0.019; 0.038]
	ARV-SD	−0.026	[−0.056; 0.003]	−0.039	[−0.070; −0.007]*
MTD Cranial	ARV	−0.019	[−0.067; 0.028]	−0.020	[−0.076; 0.036]
	MdPF	−0.032	[−0.070; 0.006]	−0.001	[−0.034; 0.033]
	ARV-SD	−0.022	[−0.049; 0.004]	−0.021	[−0.051; 0.008]
MDC Whole	ARV	0.009	[−0.016; 0.034]	0.000	[−0.024; 0.023]
	MdPF	−0.002	[−0.031; 0.027]	0.010	[−0.018; 0.038]
	ARV-SD	0.004	[−0.015; 0.023]	0.002	[−0.020; 0.024]
	SampEn	0.002	[−0.001; 0.005]	−0.002	[−0.005; 0.001]
MDA Whole	ARV	0.006	[−0.017; 0.029]	0.006	[−0.014; 0.026]
	MdPF	−0.021	[−0.056; 0.015]	−0.001	[−0.040; 0.038]
	ARV-SD	0.006	[−0.009; 0.021]	0.009	[−0.007; 0.025]
	SampEn	0.001	[−0.003; 0.004]	−0.001	[−0.003; 0.002]
MDS Whole	ARV	0.001	[−0.018; 0.020]	0.003	[−0.012; 0.018]
	MdPF	−0.021	[−0.049; 0.007]	0.000	[−0.038; 0.037]
	ARV-SD	−0.001	[−0.023; 0.020]	0.004	[−0.013; 0.022]
	SampEn	−0.001	[−0.003; 0.002]	0.000	[−0.002; 0.002]
	RPD	−0.010	[−0.021; 0.001]	−0.014	[−0.026; −0.002]*

Note: Statistical significance is accepted at $p < 0.05$ (*).

Appendix B

Table B1. Statistical results of the GEE models for the effect of task variation on the slopes of EMG parameters, corrected for the slopes of the joint angles.

	Variables		I_{break} (compared to C)		I_{work} (compared to C)		Covariate	
Muscle	Outcome	Covariate	B_1	95%-CI	B_2	95%-CI	B_3	95%-CI
MTT	ARV	Clavicle retraction	−0.013	[−0.034; 0.008]	−**0.015**↓	[−0.053; 0.023]	0.130	[−0.216; 0.477]
		Scapula protraction	−**0.003**↑	[−0.026; 0.020]	−**0.017**↓	[−0.053; 0.020]	−0.290	[−0.527; −0.053]*
		Scapula anterior tilt	−0.014	[−0.038;0.010]	−**0.018**↓	[−0.060; 0.025]	0.042	[−0.202; 0.286]
	MdPF	Clavicle retraction	**0.001**↑	[−0.055;0.057]	**0.008**↑	[−0.052; 0.067]	0.237	[−0.247; 0.721]
		Scapula protraction	−**0.006**↓	[-0.053; 0.041]	**0.000**↑	[−0.057; 0.058]	0.175	[−0.211; 0.560]
		Scapula anterior tilt	−**0.001**↓	[−0.057; 0.056]	**0.002**↑	[−0.056; 0.059]	0.029	[−0.297; 0.355]
	CoA-x	Clavicle retraction	0.000	[−0.001; 0.001]	0.000	[−0.002; 0.002]	0.006	[−0.002; 0.014]
		Scapula protraction	0.000	[−0.002; 0.001]	0.000	[−0.002; 0.002]	0.011	[0.002; 0.020]*
		Scapula anterior tilt	0.000	[−0.001; 0.002]	0.000	[−0.002; 0.002]	−0.004	[−0.014; 0.006]
	CoA-y	Clavicle retraction	0.001	[−0.000; 0.003]	0.001	[−0.001; 0.004]	−0.004	[−0.020; 0.013]
		Scapula protraction	0.001	[−0.001; 0.002]	0.001	[−0.001; 0.004]	0.017	[0.006; 0.028]*
		Scapula anterior tilt	**0.002**↑	[−0.000; 0.004]	0.001	[−0.001; 0.004]	−0.009	[−0.024; 0.005]
	ARV-SD	Clavicle retraction	−0.006	[−0.025; 0.013]	−**0.021**↓	[−0.051; 0.009]	−0.046	[−0.314; 0.223]
		Scapula protraction	−**0.003**↑	[−0.022; 0.017]	−**0.019**↓	[−0.048; 0.010]	−0.101	[−0.307; 0.105]
		Scapula anterior tilt	−**0.005**↑	[−0.023; 0.013]	−**0.020**↓	[−0.051; 0.011]	−0.039	[−0.198; 0.120]
	SampEn	Clavicle retraction	0.000	[−0.002; 0.001]	−0.001	[−0.003; 0.001]	0.006	[−0.024; 0.036]
		Scapula protraction	0.000	[−0.002; 0.001]	− −0.001	[−0.004; 0.001]	−0.005	[−0.021; 0.012]
		Scapula anterior tilt	−**0.001**↓	[−0.002; 0.001]	−0.001	[−0.004; 0.001]	0.011	[−0.005; 0.027]
MTD	ARV	Clavicle elevation	−**0.009**↑	[−0.060; 0.042]	−**0.014**↑	[−0.075; 0.046]	1.058	[0.113; 2.004]
		Scapula lateral rot.	−**0.011**↑	[−0.062; 0.041]	−**0.015**↑	[−0.076; 0.047]	0.487	[0.201; 0.772]*
		Humerus elevation	−**0.016**↑	[−0.075; 0.044]	−**0.020**↑	[−0.086; 0.047]	0.298	[0.164; 0.432]*
		Neck flexion	−**0.021**↑	[−0.071; 0.028]	−0.030	[−0.092; 0.032]	0.199	[−0.006; 0.403]
	MdPF	Clavicle elevation	−**0.020**↑	[−0.052; 0.013]	**0.020**↑	[−0.006; 0.047]	0.805	[0.248; 1.361]*
		Scapula lateral rot.	−**0.026**↑	[−0.058; 0.006]	**0.014**↑	[−0.015; 0.042]	0.198	[−0.069; 0.465]
		Humerus elevation	−**0.029**↑	[−0.058; 0.001]	**0.008**↑	[−0.013; 0.029]	0.093	[−0.022; 0.208]
		Neck flexion	−0.032	[−0.062; −0.003]*	0.005	[−0.016; 0.025]	−0.056	[−0.205; 0.094]
	CoA-x	Clavicle elevation	0.000	[−0.001; 0.001]	0.000	[−0.001; 0.001]	0.027	[0.006; 0.049]*
		Scapula lateral rot.	0.000	[−0.001; 0.001]	0.000	[−0.001; 0.001]	0.004	[−0.005; 0.013]
		Humerus elevation	0.000	[−0.001; 0.001]	0.000	[−0.001; 0.001]	−0.002	[−0.006; 0.002]
		Neck flexion	0.000	[−0.001; 0.001]	0.000	[−0.001; 0.000]	0.005	[−0.001; 0.011]
	CoA-y	Clavicle elevation	−**0.001**↓	[−0.001; 0.000]	−0.001	[−0.001; −0.000]*	−0.005	[−0.019; 0.009]
		Scapula lateral rot.	0.000	[−0.001; 0.000]	−0.001	[−0.001; −0.000]*	0.003	[−0.003; 0.009]
		Humerus elevation	−**0.001**↓	[−0.001; 0.000]	−0.001	[−0.001; 0.000]	−0.002	[−0.005; 0.002]
		Neck flexion	−**0.001**↓	[−0.001; 0.000]	−0.001	[−0.001; 0.000]	−0.005	[−0.010; 0.001]
	ARV-SD	Clavicle elevation	−**0.015**↑	[−0.043; 0.014]	−**0.017**↑	[−0.050; 0.015]	0.646	[0.094; 1.198]*
		Scapula lateral rot.	−**0.018**↑	[−0.046; 0.011]	−**0.020**↑	[−0.050; 0.010]	0.237	[0.077; 0.396]*
		Humerus elevation	−**0.022**↑	[−0.053; 0.009]	−0.026	[−0.058; 0.007]	0.079	[−0.022; 0.179]
		Neck flexion	−0.023	[−0.050; 0.004]	−0.028	[−0.058; 0.002]	0.073	[−0.100; 0.246]
	SampEn	Clavicle elevation	**0.000**↑	[−0.003; 0.003]	−**0.001**↑	[−0.004; 0.002]	0.050	[−0.008; 0.109]
		Scapula lateral rot.	**0.000**↑	[−0.003; 0.002]	−**0.001**↑	[−0.004; 0.002]	0.019	[−0.005; 0.043]
		Humerus elevation	−0.001	[−0.004; 0.002]	−**0.001**↑	[−0.005; 0.002]	0.007	[−0.005; 0.020]

		Neck flexion	−0.001	[−0.004; 0.002]	−0.002	[−0.005; 0.002]	−0.001	[−0.017; 0.014]
MDC	ARV	Humerus flexion	0.009	[−0.015; 0.033]	**−0.001** ↓	[−0.023; 0.021]	0.002	[−0.082; 0.087]
		Humerus elevation	**0.012** ↑	[−0.013; 0.037]	**0.003** ↑	[−0.026; 0.031]	0.088	[−0.087; 0.263]
	MdPF	Humerus flexion	**−0.009** ↓	[−0.036; 0.018]	**0.001** ↓	[−0.028; 0.030]	0.070	[−0.016; 0.157]
		Humerus elevation	**0.003** ↑	[−0.021; 0.027]	**0.015** ↑	[−0.004; 0.035]	0.157	[0.056; 0.257]*
	ARV-SD	Humerus flexion	**0.007** ↑	[−0.008; 0.022]	**0.006** ↑	[−0.009; 0.020]	−0.028	[−0.122; 0.065]
		Humerus elevation	**0.005** ↑	[−0.016; 0.026]	**0.003** ↑	[−0.024; 0.030]	0.029	[−0.142; 0.200]
	SampEn	Humerus flexion	0.002	[−0.001; 0.005]	−0.002	[−0.004; 0.001]	−0.002	[−0.017; 0.012]
		Humerus elevation	**0.003** ↑	[−0.001; 0.006]	**−0.001** ↑	[−0.005; 0.002]	0.020	[0.008; 0.033]*
MDA	ARV	Humerus flexion	**0.001** ↓	[−0.018; 0.019]	**−0.000** ↓	[−0.019; 0.019]	0.046	[−0.034; 0.126]
		Humerus elevation	**0.013** ↑	[−0.002; 0.028]	**0.015** ↑	[−0.005; 0.035]	0.250	[0.124; 0.376]*
	MdPF	Humerus flexion	**−0.012** ↑	[−0.042; 0.017]	**0.009** ↑	[−0.038; 0.056]	−0.077	[−0.251; 0.097]
		Humerus elevation	**−0.008** ↑	[−0.031; 0.015]	**0.014** ↑	[−0.015; 0.043]	0.423	[0.260; 0.586]*
	ARV-SD	Humerus flexion	**0.001** ↓	[−0.011; 0.013]	**0.003** ↓	[−0.013; 0.020]	0.043	[−0.001; 0.087]
		Humerus elevation	**0.011** ↑	[0.001; 0.021]*	**0.015** ↑	[−0.002; 0.033]	0.185	[0.074; 0.295]*
	SampEn	Humerus flexion	0.001	[−0.002; 0.004]	**0.000** ↑	[−0.002; 0.001]	−0.002	[−0.011; 0.008]
		Humerus elevation	0.001	[−0.002; 0.005]	**0.000** ↑	[−0.002; 0.002]	0.015	[0.003; 0.026]*
MDS	ARV	Humerus flexion	**0.002** ↑	[−0.013; 0.018]	**0.004** ↑	[−0.011; 0.020]	−0.010	[−0.084; 0.063]
		Humerus elevation	**0.007** ↑	[−0.009; 0.022]	**0.010** ↑	[−0.004; 0.023]	0.191	[0.146; 0.235]*
	MdPF	Humerus flexion	**−0.019** ↑	[−0.054; 0.016]	**0.002** ↑	[−0.042; 0.045]	−0.017	[−0.148; 0.114]
		Humerus elevation	**−0.016** ↑	[−0.044; 0.012]	**0.005** ↑	[−0.035; 0.046]	0.167	[0.032; 0.303]*
	ARV-SD	Humerus flexion	**0.000** ↑	[−0.020; 0.020]	**0.005** ↑	[−0.013; 0.024]	−0.009	[−0.078; 0.060]
		Humerus elevation	**0.004** ↑	[−0.015; 0.024]	**0.011** ↑	[−0.008; 0.030]	0.198	[0.147; 0.248]*
	SampEn	Humerus flexion	**0.000** ↑	[−0.002; 0.002]	**0.001** ↑	[−0.001; 0.002]	−0.007	[−0.018; 0.003]
		Humerus elevation	**0.000** ↑	[−0.003; 0.002]	0.000	[−0.002; 0.002]	0.012	[0.006; 0.019]*

Notes: When the covariate (slope of joint angle) creates a change of >10% for the B's of I_{break} or I_{work} versus the continuous condition (C), the B's are in bold. Statistical significance is accepted at $p < 0.05$ (*).