Effects of additional weight on posture-movement adaptations to repetitive arm motion-induced fatigue

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For ‘Tía Tomy’ and ‘Tío Javier’

CONTRIBUTION OF AUTHORS

Hiram A. Cantú, the candidate, was responsible for the research, design, setup, recruitment, data collection, analysis, writing and any other step related to the completion of the research study and the submission of this thesis as per McGill University requirements.

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Kim Emery, MSc, assisted with the training of the candidate and provided guidance during the data collection and analysis.

Karen Lomond, PhD, and David Antle, PhD candidate, provided assistance during data processing and analysis.

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ABSTRACT

The aim of this Master's study was to quantify the effect of adding weight to the trunk on posture-movement adaptations in healthy young adults during repetitive arm motion-induced fatigue. Whole-body kinematic, kinetic and electromyographic (EMG) data were recorded during a repetitive pointing task (RPT) in normal conditions and with a 20% body weight extra load. The first and last minutes of the RPT were analyzed for both sessions. Results show that the RPT was effective in inducing muscular fatigue in three of the four upper limb muscles investigated. Other results of this study confirm the ability of the body system to adapt to fatigue in different postural conditions. A few different adaptations to fatigue were selected by the system in the added weight condition to contribute to the performance of the task and reach a similar time to fatigue as in normal condition. Results suggest that adaptations to fatigue in the extra weight condition could aim at ensuring postural stability by maintaining the body's center of mass stable, in addition to reducing the load of the fatigued musculature.

ABRÉGÉ

L'objectif de cette étude de maitrise était de quantifier les effets d'ajouter du poids sur le tronc corporel sur les adaptations de la posture et du mouvement de jeunes adultes en santé durant la fatigue induite par des mouvements répétitifs du bras. Des données cinématiques, cinétiques et électromyographiques (EMG) ont été enregistrées durant une tâche de pointage répétitif (TPR) en condition normale et avec une charge additionnelle de 20% du poids corporel. Les premières et dernières minutes de la TRP ont été analysées pour les deux séances. Les résultats démontrent que la TRP a été efficace à induire de la fatigue musculaire à trois des quatre muscles du membre supérieur. Les autres résultats de cette étude confirment la capacité du système corporel à s'adapter à la fatigue dans différentes conditions posturales. Quelques adaptations différentes à la fatigue ont été sélectionnées par le système dans la condition de charge additionnelle afin de contribuer à la performance de la tâche et d'atteindre un temps total de tâche similaire à celui en condition normale. Les résultats suggèrent que les adaptations à la fatigue et à la charge additionnelle pourraient viser à assurer la stabilité posturale en maintenant un centre de masse corporel stable tout en réduisant la charge sur les muscles fatigués.

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INTRODUCTION

The main function of the postural control system is to stabilize posture against gravity and therefore to maintain balance (Massion, 1994). In order to maintain balance, equilibrium between the center of mass (CoM) of the body and the center of pressure (CoP) under the feet must exist. CoM is the point in the body around which its weight is balanced and it is located below the navel, around 55-57% of the standing height (McGinnis, 2005). In biomechanics, CoP refers to the point where all the ground reaction forces act (McGinnis, 2005). Winter and colleagues generated a model that is considered the most reliable in explaining postural control of the human body and is known as the inverted pendulum model. This model establishes that the CoP movements seen under the feet follow and counteract the CoM movements while the body sways, suggesting the role of the CoP as a control variable and the CoM as the controlled variable. In this model, the postural system aims to keep the spatial difference between the vertical projection of the CoM and of the CoP as small as possible, resulting in a stable posture (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998).

Despite this role that has been previously assigned to the CoP, some recent studies challenge its predictions. After blocking the trunk of the body of participants, Carpenter and colleagues found that CoM displacements were minimized but CoP showed an unexpected increase of its displacement (Carpenter, Murnaghan, & Inglis, 2010). In addition, more recent experiments found that when fatigue was induced in one leg, the increments of CoP displacements were only present under the non-fatigued leg. Similar results were seen on patients with unilateral hip arthroplasty, where the increments of CoP displacements were seen only under the sound leg (Belaid et al., 2007). Taken together, these recent studies suggest that the control of the CoP could be used for strategies other than maintaining the best alignment possible with the CoM, at least in some circumstances.

Recently, some authors have focused their efforts on studying postural control using dual task paradigms, in many cases with additional tasks that are accomplished in everyday life. During an arm-raising task from a standing position, it was found that if the rate of velocity of the arm-raising task increases, the amplitude of the CoP displacements also increases (Ferry, Martin, Termoz, Cote, & Prince, 2004). Another similar study investigated the arm-trunk coordination when reaching beyond arm's length. The expectation from the authors was that there would be a decrease in the CoM displacement in order to compensate for the postural perturbations caused by the arm movements, but the results rather showed that the CoM accelerates towards the place where the arm is displaced and the CoP tracks these accelerations (Pozzo, Stapley, & Papaxanthis, 2002). Together, these studies suggest that when postural control is combined with another task, predictions from classical postural control models may not apply.

In everyday tasks where standing postural control is combined with arm tasks, the arm tasks are often performed in a repetitive fashion. Repetitive arm movements can be performed in many activities of daily life such as during work or in leisure activities (Zakaria, Robertson, MacDermid, Hartford, & Koval, 2002). Repetitive motion is also a factor associated with the development of musculoskeletal problems by way of muscle fatigue. Fatigue has been defined as "an acute impairment in performance that includes both an increase in the perceived effort necessary to exert a desired force and an eventual inability to produce this force" (Enoka & Stuart, 1992). Statistics show fatigue as a potential risk factor of injuries by falls (Parijat & Lockhart, 2008). In addition, in Canada, in the last decade, 55% of injuries caused by a repetitive motion occurred at the workplace and 25% affected the neck and shoulder area (Tjepkema, 2003).

Kinetic, kinematic and electromyographic characteristics seem to be affected when fatigue is induced by a repetitive task. Authors found increases in postural sway whether fatigue is induced directly in postural muscles by repeated plantar-flexion exercises or it is induced in the upper limb after a repetitive arm movement (Corbeil, Blouin, Bégin, Nougier, & Teasdale, 2003; Nussbaum, 2003). In addition, when shoulder fatigue is induced by repetitive arm movements, posture-movement adaptations are observed and are interpreted as a compensation mechanism to reduce the load of the fatigued musculature and to continue with the performance of the task (Fuller, Lomond, Fung, & Côté, 2009). These adaptations include electromyographic and kinematic changes indicative of fatigue at the upper limb (e.g. increased upper trapezius EMG activity amplitude) as well as increased involvement of the trunk to facilitate movement of the terminal segment (e.g. hand) (Côté, Feldman, Mathieu, & Levin, 2008; Côté, Mathieu, Levin, & Feldman, 2002; Côté, Raymond, Mathieu, Feldman, & Levin, 2005). However, many aspects of the combined control of standing posture and movement during repetitive arm motion-induced fatigue remain poorly understood.

Many other factors have an effect on postural control that could end in an injury. Hue and colleagues showed that adding weight to the human body is strongly correlated with a decrease in posture stability (Hue et al., 2007). Effects of fatigue and of extra body weight applied to the body on postural control have been compared, resulting in larger body sway for both conditions but more pronounced with additional body weight than with fatigue (Ledin, Fransson, & Magnusson, 2004). However, to our knowledge, these two factors have never been studied together during repetitive upper limb movements. Where increases in postural sway and posture-movement adaptations are found with fatigue induced by a repetitive arm task, it remains unknown whether the same strategies will be employed by the system if posture stability is challenged (e.g. by adding weight to the body) during an upper-limb task or whether different posture-movement adaptations will emerge due to the postural threat.

In summary, the whole-body changes previously observed to occur after a repetitive movement have been interpreted as voluntary adaptations where the trunk is recruited to contribute to the arm sub-task. This strategy of the trunk helping the fatigued limb to move from one place to the other is supported by previous studies on impairment or loss of somatosensory information whereby the body increased CoP displacements as a strategy to compensate for feedback lost from the conditions (Belaid, et al., 2007; Carpenter, et al., 2010; Vuillerme, Sporbert, & Pinsault, 2009). One-way of challenging this hypothesis is by further challenging posture. In the current study, we added weight to the trunk of the participants to destabilize the trunk movements and jeopardize postural stability, so that it may not be as much of a good strategy to use posture-movement adaptations that help the performance of the arm sub-task. In that case, we hypothesized that posture-movement adaptations described previously would be replaced by other strategies not as threatening to posture, illustrating the ability of the system to take advantage of its redundancy in finding alternative solutions to a challenging task. Thus, the main objective of this Master's thesis was to quantify the effect of adding weight to the trunk on the posture-movement adaptations during repetitive arm motion-induced shoulder fatigue.

The findings of this study could provide important knowledge to better understand postural control and how the body coordinates posture and movement. In addition, knowing how the body adapts to scenarios where fatigue is induced after a repetitive task and/or extra body weight is added to the trunk of the body, may help us develop better interventions for the prevention of injuries in athletes, workers who perform repetitive tasks and for the management of injuries or diseases linked to fatigue and postural control, such as diabetes and obesity, among others.

LITERATURE REVIEW

Postural Control Models

The central nervous system is considered responsible for coordinating the activities of the whole body. One such important activity is postural stability, for which equilibrium between center of mass (CoM) of the body and center of pressure (CoP) under the base of support must exist. Since the body's CoM is in constant movement due to many factors such as breathing (Jeong, 1991) and muscle activity (Soames & Atha, 1981, 1982), the CoP is in constant need to track the position of the CoM in order to maintain balance. Literature suggests that this reaction is made by feedback received from multiple sources of sensory information, with vision and the vestibular system as two of the most important feedback means (Ishida & Miyazaki, 1987; Johansson, Magnusson, & Akesson, 1988; Massion, 1994; Peterka, 2000).

Many models have been proposed to explain postural control of the human body. Human bipedal stance is considered unstable and quiet standing requires continuous postural stabilization. Winter and colleagues (1998) proposed a model whereby the central nervous system (CNS) maintains postural control. Authors established that the body acts as an inverted pendulum, which is a relatively simple control scheme that regulates posture with quick responses that reduce the operating demand on the CNS. In their model, the muscles act as spring that control stiffness and where the CoP moves in phase with the CoM as the body sways in two planes, the sagittal (along the anterior-posterior axis) and the frontal (along the medial-lateral axis) ones. This general theory is supported by many other studies (Rougier, 2009; Rougier & Bergeau, 2009). In the anterior-posterior (AP) direction, the ankle plantarflexor and dorsiflexor torques are thought to control the CoP, while in the medial-lateral (ML) direction, the CoP is thought to be controlled by the hip abductor and adductor torque (Rougier, 2007; Winter, et al., 1998). The combination of control processes in these two axes helps to optimize postural performance (Winter, Prince, Frank, Powell, & Zabjek, 1996). These studies also established that the CoP movements are observed to follow and counteract the movements of the CoM while the body sways, suggesting that the CoP could be a control variable and the CoM a controlled variable (Corriveau, Hebert, Prince, & Raiche, 2000).

After several and relevant tests made by Winter and colleagues to their own proposed model, authors found quantitative evidence that the CoP tracks the CoM, likely to keep it within a desired position between the two feet, supporting this principle of postural control. However, the principal measure that validates the model is the correlation between the CoM acceleration and the CoP-CoM (error signal as the CoP tracks the CoM) or the two-dimensional difference in distance between the location of CoP and the vertical projection of the CoM (Corriveau, et al., 2000). Results showed that the CoP oscillates in phase with the CoM with small negative lag time, which is interpreted to mean that the model is a slightly damped system.

Even though the inverted pendulum model proposed by Winter has been accepted as the most appropriate model explaining how postural control of the human body works, other models have been presented in recent years. Notably, one model emphasizes multisensory postural control and suggests four sensory signals as the essential minimum elements involved in postural control: (1) an otolith-derived signal of the body lean in space, which is believed to be at the origin of the parts of the vestibular information that are the most relevant for postural control; (2) a canal-derived signal of support surface motion in space, with helps humans to perceive rotation of the support surface; (3) somatosensory graviception from deep plantar mechanoreceptors, since evidence shows that postural control is impaired with the loss of plantar somatosensory; and (4) a proprioceptive signal of body angular position with respect to the foot (Mergner, Maurer, & Peterka, 2003). This and other recent models provide a more unified point of view between the biomechanics and the neuromotor aspects of how we may control posture.

Despite the role of control variable previously assigned to the CoP to ensure postural stability, recent findings challenge this theory. Carpenter and colleagues (2010) found an increase in CoP displacement when the CoM displacement was minimized by blocking the trunk of the body. Moreover, when fatigue was induced in one leg, the increments of CoP displacements were only present under the non-fatigued leg (Vuillerme, et al., 2009). Similar findings were seen in patients with unilateral hip arthroplasty, where the increments of CoP displacements were seen under the sound leg (Belaid, et al., 2007). Together, these studies suggest that at least in some cases, CoP might have roles other than to simply follow and maintain CoM within a safe zone. Other proposed objectives pursued by moving the CoP are to provide adequate feedback about the state of the postural system especially in situations when it is disrupted from its normal state such as in fatigue or injury (Vuillerme, et al., 2009), which is in line with the model proposed by Mergner et al. (2003).

Coordination Between Posture and Movements

Different experimental paradigms have been used to study postural control. One of the most common is to study postural control during the performance of a second task (Kang & Lipsitz, 2010). A common such type of dual task is arm motion. During quiet standing and with the performance of an arm-raising task, changes in the strategies to maintain balance are seen at the hip, while at the ankle, measurements indicate strategies to modify torque and control CoP (Ferry, et al., 2004). In the same study, an increase in the amplitude of the CoP displacement was seen when the rate of velocity of the performance of the arm-raising task increased. In another study combining postural control and arm motion as a dual task, it was seen that the CoM accelerates towards the place where the arm is displaced and the CoP tracks these accelerations (Pozzo, et al., 2002). Moreover, in this study, it was observed that the CoM moved in phase with the arm during a task of reaching beyond the immediate workspace, suggesting that when necessary, the CoM could be recruited to contribute to the main arm task. Together, these findings support the idea that regardless of the objective pursued by the CoM (i.e. maintain a stable posture or contribute to limb movements), the system may achieve the desired movements of the CoM by controlling the CoP. In turn, the system may respond to the movements of the CoM by changing the CoP and maintaining the body in a position that is as balanced as the task affords, since the more distance there is between the CoP and CoM, the more unstable the human body is (Ferry, et al., 2004).

Another dual task paradigm that is frequently used is the combination of postural maintenance and the performance of a mental exercise. Using the inverted pendulum as the postural model, Kang and Lipsitz analyzed the stiffness control of balance in older adults during quiet standing while performing a cognitive task of serial subtractions (2010); the dual task performance brought increases in postural sway in the ML axis, but not in the AP, with a more pronounced increase in the subjects who showed more difficulty in calculating the subtractions, suggesting greater vulnerability to accidents by falls (Kang & Lipsitz, 2010). Conversely, another study showed a decrease in CoP displacement in the AP direction during bipedal quiet standing combined with the performance of a challenging mental arithmetic task (Vuillerme & Vincent, 2006). Authors suggested that the decreases in AP CoP displacements might be associated with increased stiffness and a reduction of the exploratory behavior in this direction. Taken together, these results suggest that the system responds differently in the AP and ML directions to postural disturbances brought about by a secondary task. It is suggested that the increase in postural sway in the ML direction but not in the AP direction reflects that the system benefits from a comparatively larger base of support in the ML direction and may thus choose to prioritize stabilization strategies in the direction where the size of the base of support is the smallest, i.e. the AP direction (Kang & Lipsitz, 2010). This suggests a direction-dependent ability of the postural control system that may be related to the mechanical characteristics of postural stability in both AP and ML directions.

Fatigue

A common perturbation to the posture and movement control system is fatigue. Fatigue has been defined as "an acute impairment in performance that includes both an increase in the perceived effort necessary to exert a desired force and an eventual inability to produce this force" (Enoka & Stuart, 1992). Dugan and Frontera (2000) have identified a series of potential fatigue mechanisms within the muscle during its normal force production. These mechanisms are generally classified into two categories, those relating primarily to central factors and those that are predominantly peripheral in origin. Central causes of fatigue include neural components like those related to the CNS and behavioral components such as motivation. Peripheral causes include those related to the peripheral nervous system through to the muscle cell, including: excitation-contraction coupling, energy supply, and the force generation process. Different observations are documented for fatigue during different types of tasks. For instance, a decline in force output is documented when fatigue is induced by sustained maximal contractions (Bigland-Ritchie, Johansson, Lippold, & Woods, 1983), whereas an increase in perceived effort and a reduction in the maximal force generating capacity is seen in low force tasks (Vollestad, 1997).

To better understand the nature of muscle fatigue, different measures have been developed by researchers, with this review focusing on non-invasive measures. The surface electromyogram (sEMG) is the electrical representation of the neuromuscular activation associated with muscle contraction in superficial muscles (Vollestad, 1997) and has been used as an indirect measure of fatigue for several decades (Bigland-Ritchie, et al., 1983; Enoka & Stuart, 1992). The root-mean-squared amplitude represents the power of the signal and is easily calculated and commonly used. The changes in the sEMG frequency spectrum associated with fatigue are well documented (Bigland-Ritchie, et al., 1983). During maximal isometric contractions, sEMG amplitude declines have been reported to occur with fatigue (Bigland-Ritchie, Donovan, & Roussos, 1981; De Luca, 1984), while during repetitive and sustained submaximal contractions increases in amplitude are documented (Fallentin, Jorgensen, & Simonsen, 1993; Krogh-Lund & Jorgensen, 1992). Endurance time is a basic indicator of muscle fatigue and represents the duration that a task can be performed until exhaustion (when force or intensity can no longer be maintained). Rating of perceived exertion has been used also as a basic tool to assess muscle fatigue; as such, the Borg CR-10 scale is a rating scale ranging from 0 to 10 that has been validated as a reliable method of assessing fatigue (Borg, 1982; Jones & Hunter, 1983). However, it is recommended that the last two methods be combined with additional measures since endurance time may be affected by mechanisms other than force generation and perceived exertion does not provide insight into the specific fatigue mechanisms.

Fatigue has been identified as a potential factor that affects postural control and increases injuries by falls. Previous studies show that when fatigue is induced directly in the ankle plantar-flexor muscle with repeated plantar-flexion of both legs, it results in increased postural sway (Corbeil, et al., 2003). Added to these results, it was found that when fatigue is induced after a repetitive arm movement, increases of postural sway are also present during quiet standing (Nussbaum, 2003). Thus, the literature suggests that whether fatigue is induced in muscles directly responsible for maintaining the standing posture or in other ones, the consequence is a decrease in postural stability.

Fatigue not only affects postural control, but also results in kinematic changes that have been documented. In a series of studies investigating the effects of fatigue induced by repetitive hammering and sawing, Côté et al. (Côté, et al., 2002; Côté, et al., 2005) have shown that smaller motion amplitude at the elbow was compensated by increased trunk motion during both hammering and sawing. In another study on hammering, it was observed that motion adaptations were accompanied by an increased EMG activity in upper limb musculature (trapezius) as well as in muscles distant to the area of fatigue (external oblique) (Côté, et al., 2008). These findings provided not only evidence that changes in inter-muscular coordination occur with fatigue, but also suggested that motion adaptations may occur across more than one of the planes since the external oblique is both a trunk flexor and trunk rotator. Finally, Fuller and colleagues (2009) studied the effects of repetitive pointing movements in a horizontal plane at shoulder height and showed that posture and movement adaptations occurred in all three planes of motion and included several changes that could affect postural stability. Such changes included lateral shifts of the body towards the contralateral (non-moving) arm and increases in AP CoM and CoP movement amplitudes, in phase with movements of the arm. Taken together, these findings suggest that posture-movement adaptations occur as a compensation mechanism when shoulder fatigue is present after a repetitive movement (Fuller, et al., 2009). As has been suggested in studies by Pozzo and colleagues, these previous findings suggest that in situations such as arm fatigue, the system is able to change the role of the trunk from principally stabilizing posture to one of contributing to the movements of the fatigued arm. However, these compensations may have detrimental consequences on postural stability, since lateral postural shifts and increased AP movements may move the CoP closer to the edges of the standing base of support and may pose a threat that the system may want to prevent, as suggested by Kang and Lipsitz (2010). Thus, upper limb fatigue offers a model to study how posture and movements are coordinated in successfully accomplishing tasks of daily life.

Body Weight as a Predictor of Postural Stability

Adding weight to the human body has been identified as another factor that affects postural control, with a decrease of postural stability that may result in an injury. Few authors have studied how additional weight can affect posture. Hue and colleagues aimed to determine the contribution of body weight on postural stability in conditions where the subjects were tested with vision and without it. In both cases, body weight accounted for a big proportion of the balance strategies, suggesting that body weight may be considered as a risk factor for falls (Hue, et al., 2007). Other studies showed that body sway is more affected by additional weight in comparison with muscle fatigue. Ledin et al. (2004) exposed their participants to vibratory proprioceptive stimulation in normal condition, when the body weight was increased by 20% and when the triceps surae muscles were fatigued. They found that postural control was affected by both additional body weight and muscle fatigue, with a larger body sway in both cases. However body sway was significantly larger in the extra body weight condition compared to the conditions where fatigue was induced. Similar effects of additional body weight and fatigue were found on gait characteristics, increasing hip and trunk range of motion (Qu & Yeo, 2011). Taken together, the literature suggests that increasing weight of the body may not only result in a decrease in gait performance, but also in an increase in fall risk (Park, Hur, Rosengren, Horn, & Hsiao-Wecksler, 2010). Moreover, obesity has been identified as a threat to balance control. Berrigan and colleagues (2006) found greater CoP displacement in obese patients than in normal individuals while standing and performing a pointing task. After obese patients lost weight, reductions in CoP range of motion were seen in all directions (Teasdale et al., 2007), further reinforcing the belief that extra body weight represents a postural threat.

In another study, three different loads were applied to the participants, with results showing that with an increase in body weight, the random movements of the postural sway decreased, the CoP displacements increased linearly and there was not a significant change in muscle activity (Schiffman, Bensel, Hasselquist, Gregorczyk, & Piscitelle, 2006), suggesting a direct relationship between the level of threat and the intensity of the postural stabilization response. Finally, changes in postural angles were analyzed in a study on preadolescent children fitted with backpacks containing various loads. Results showed that carrying a load weighing 15% of the total bodyweight significantly affected all of the measured postural angles of the participants (Ramprasad, Alias, & Raghuveer, 2010). Taken together, these results suggest that postural disturbances exist after a load weight is added to the body, resulting in adaptations of the system that might be voluntary strategies to reduce the threat (i.e. by reducing postural sway) and/or to provide feedback for any mishap that may arise (i.e. by increasing CoP displacement). Despite this, no study thus far has measured the combined effect of added weight and arm fatigue on the posture and movement strategies developed to maintain the performance of an arm task accomplished from a standing position.

RESEARCH ARTICLE

Effects of additional weight on posture-movement adaptations to repetitive arm motion-induced fatigue

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ABSTRACT

Fatigue and extra body weight have been identified as two separate factors that can decrease postural stability and increase risk of injuries by falls. We have previously shown that posture and movement adaptations occur in all planes of motion when people accomplish repetitive upper limb tasks to fatigue while standing, and act as strategies to reduce the load on the fatigued arm musculature. These notably include lateral body shifts towards the contralateral (non-moving arm) side and increased trunk range of motion along with the arm movement. While these strategies facilitate the arm movement sub-task, they may jeopardize postural stability and represent a threat to equilibrium, especially if additional weight is placed on the trunk. The goal of this study was to assess the effects of adding weight to the trunk on posture-movement adaptations in healthy young adults during repetitive arm motion-induced fatigue. A sample of 19 healthy young adults was recruited to perform two sessions, assigned in random order, of the same repetitive pointing task (RPT) from a standing position: one in normal conditions (No-Belt) and another with an added load of 20% body weight around the waist (Belt). Whole-body kinematic, kinetic and EMG characteristics were recorded and the first and last minutes of the RPT (Time: No-fatigue and Fatigue-terminal, respectively) were analyzed for both sessions. EMG amplitude in both conditions significantly increased from the first to the last minute of the RPT in the anterior deltoid (18.84 ± 28.24 %), biceps (35.72 ± 32.65 %) and upper trapezius (23.99 ± 24.03 %) indicating that arm fatigue was effectively induced. There was a significant Belt by Time interaction effect on peak center of pressure (CoP) velocity in the medial-lateral direction ($p < 0.04$). Regardless of fatigue, the maximum CoP velocity was lower for the added weight condition, and fatigue only affected this parameter in normal conditions by reducing it. There was also a significant Belt by Time interaction effect on the reaching shoulder's average vertical position ($p = 0.01$). Regardless of time, it was lower with the added load and its increase with fatigue was more pronounced in the No-Belt condition. As described in previous studies, the reaching shoulder and elbow

joints were shifted laterally towards the non-reaching side. However, the lateral shifts seen before in the center of mass (CoM) and CoP were not present in this study. Increases in ranges of motion (ROMs) were found in all the kinematic parameters of interest except for the body's CoM. These findings suggest that with these few exceptions, fatigue adaptations are robust across different postural conditions. While supporting the previous interpretation that these adaptations reflect voluntary strategies of the system to contribute to the arm sub-task, the current study suggests that especially with additional postural constraints, they also ensure postural stability by maintaining the CoM constant within a stable range of motion.

1. Introduction

The main function of the postural control system of the human body is to maintain balance (Massion, 1994). To achieve this while standing, the average point of application of ground reaction forces, also known as center of pressure (CoP), follows and counteracts the movements of the body's center of mass (CoM) (Winter, et al., 1998). According to this, one expects increased movements of the CoM to be followed by increases in those of the CoP. However, recent studies question this prediction. In an experiment where CoM movements were minimized by blocking the trunk of the body, CoP displacements rather increased (Carpenter, et al., 2010). In line with this result, when fatigue was induced in one leg, the increments of CoP displacements were only present under the non-fatigued leg as it was also seen in patients with unilateral hip arthroplasty, where the increments of CoP displacements were only under the sound leg (Belaid, et al., 2007; Vuillerme, et al., 2009). Together, these studies suggest that the goal of controlling the CoP may not only be to passively follow the CoM but may serve to provide a mechanism to obtain postural feedback in situations where access to postural information is challenged or reduced, as suggested by Vuillerme and colleagues (2009).

When standing, we often perform tasks with our upper limbs during daily activities, oftentimes in a repetitive fashion (Zakaria, et al., 2002). Repetitive arm movements such as those perform at work generally lead to a reduced functional capacity of the working arm muscles, which is better known as fatigue. Of the total injuries caused by a repetitive motion in Canada, 55% occurred at the workplace and 25% affected the neck and shoulder area (Tjepkema, 2003). In addition, statistics show fatigue as a potential risk factor of injuries by falls (Parijat & Lockhart, 2008). Increases of postural sway have been documented when fatigue is induced directly in the ankle plantar-flexor muscle (Corbeil, et al., 2003) and when it is induced after a repetitive arm movement (Nussbaum, 2003). In addition, posture-movement adaptations have been observed as a result of repetitive upper limb motion-induced fatigue and have been interpreted as a compensation mechanism to reduce the load on the fatigued musculature (Fuller, et al., 2009). Indeed, work by the group of Côté et al. showed that during various repetitive upper limb tasks, signs of localized arm fatigue were accompanied by increased contribution of trunk muscles and increased trunk movements in the direction of the arm movement (Côté, et al., 2008; Côté, et al., 2002; Côté, et al., 2005). Postural shifts towards the non-moving arm side were also observed; taken together, these studies suggest that postural compensatory strategies aimed to contribute to the arm sub-task but may at the same time jeopardize postural stability.

Adding weight to the trunk of the body is also strongly correlated with a decrease in posture stability. Hue and colleagues showed that with and without vision, body weight was one of the most important factors affecting postural stability (Hue, et al., 2007). Another study compared the effects of lower limb fatigue and extra body weight applied to the body on postural control and larger body sway was observed for both conditions but was more pronounced with additional body weight (Ledin, et al., 2004). Moreover, a correlation is also seen between obesity and balance control. Greater CoP displacement was found in obese patients compared to normal individuals while standing and performing a task (Berrigan, et al., 2006). Teasdale and colleagues (2007) reported reduction in CoP range of motion in all directions after obese patients lost weight.

In summary, previously described postural adaptations to repetitive arm motion-induced fatigue may jeopardize postural stability for the benefit of contributing to the fatigued arm's sub-task. It is unknown to what extent the postural system would develop strategies that jeopardize postural stability if postural constraints were more important, for instance in the presence of extra trunk weight. Therefore, the aim of the current work was to quantify the effect of adding weight to the trunk of the body on the posture-movement adaptations to repetitive arm motion-induced shoulder fatigue. We hypothesized that with extra body weight, strategies other than those decreasing postural stability would be developed to compensate for arm muscle fatigue.

2. Methods

2.1. Participants

A convenience sample of 19 healthy young adults (10 men, 9 women; mean age = 28.58 ± 6.24 years; mean height = 169.53 ± 7.43 cm; mean mass = 65.41 ± 8.69 kg; mean BMI = 22.69 ± 2.02 kg/m^2) was recruited through personal contacts to participate in this study. Subjects were excluded if they had any neuromusculoskeletal or cardiovascular impairment or diagnosed condition that could affect the performance of the experiment. All subjects were right-handed. The study was performed at the Occupational Biomechanics and Ergonomics Lab (OBEL) of the Jewish Rehabilitation Hospital in Laval, Quebec. At arrival, subjects provided written informed consent prior to participation by signing forms approved by the Research Ethics Board of the Centre for Interdisciplinary Research in Rehabilitation (CRIR) of Greater Montreal. All participants performed the fatigue protocol twice (see below) with a mean of 7.63 ± 3.52 days between both sessions. All subjects were asked to refrain from changing their usual physical activity routine between the two testing sessions.

2.2. Experimental Protocol

The experiment conducted was a repeated measures design. In one of the two sessions, subjects performed the fatigue protocol in normal conditions, while in the other they performed the protocol with an added load of 20% of their body weight. This was accomplished by having participants wear a belt (MiR Champion Belt, MiR Vest, Inc.), placed at waist height and secured closely but comfortably around the waist. The belt was filled with removable weight blocks for a maximum load of 36 pounds. Individual 1.5 pounds weight blocks were spaced as evenly around the waist as possible, secured by the belt and attached with velcro straps. The order of the sessions was randomized for every subject.

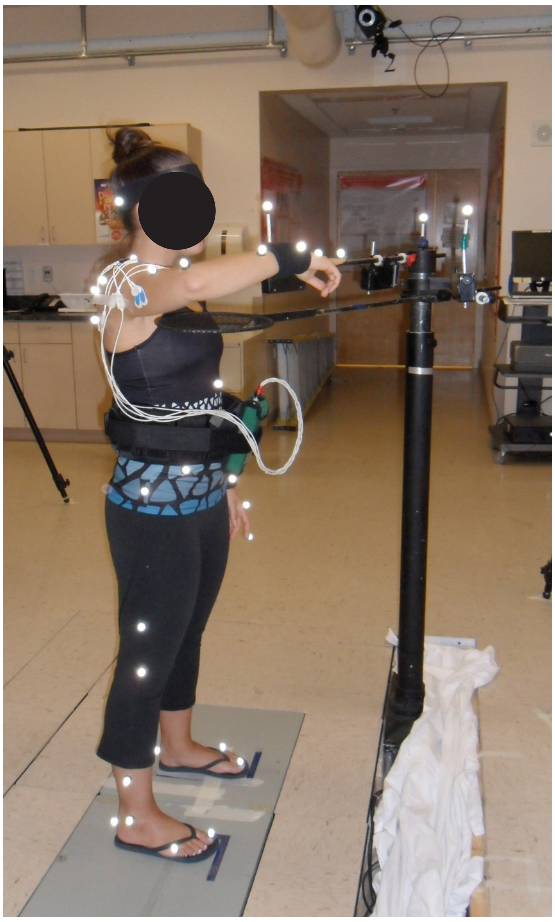

Fig. 1. Experimental setup of a subject wearing the belt.

The choice of the task for the experimental protocol was largely based on previous studies conducted in the OBEL lab, where it has been confirmed that fatigue is induced in the upper limb after the repetitive task is performed (Emery & Côté, 2012; Fuller, et al., 2009; Lomond & Côté, 2011). The protocol consisted of performing a repetitive pointing task (RPT) with the dominant arm to fatigue.

To guide the RPT, two cylindrical touch-sensitive targets (length: 6 cm, radius: 0.5 cm, response time: 130 ms, Quantum Research Group Ltd.) were placed in front of the subject's midline, at shoulder height, at 100% (distal target) and 30% (proximal target) of arm length. In addition, an elliptically shaped mesh barrier (major axis: 24.5 cm, minor axis: 20.5 cm) was placed under the elbow joint's functional range of motion to ensure that the entire arm moved in a horizontal plane at shoulder height for the entire duration of the RPT. The position of the barrier did not restrict the natural trunk movement.

For the RPT, participants stood over two adjacent force platform, 25.4 cm of distance between feet, and reached back and forth between the two targets with their dominant arm using the index finger. They touched each target while keeping the elbow at shoulder height and without touching the barrier. Their left arm rested on the side of the body during the whole RPT. Subjects were to maintain a rhythm of one movement per second with the help of a metronome that was set up with a frequency of 1 Hz. Since the targets emitted a sound when touched, subjects were to match the sounds of the metronome to those of the targets as closely as possible. Subjects were asked to perform the task as naturally and as long as possible.

During the last 30 s of every minute that the task was performed, biomechanical data was recorded, at the end of which subjects were to provide their rating of self-perceived exertion in shoulder and neck region with the help of a Borg CR-10 scale (Borg, 1982). Four stoppage criteria were used: (1) subjects reached a perceived level of exertion of eight for the shoulder and/or neck region on the Borg CR-10 scale, (2) subjects felt that they could not continue the RPT, (3) the elbow was touching the barrier or (4) subjects could no longer maintain the 1 Hz movement rhythm (Côté, et al., 2002; Fuller, et al., 2009; Hammarskjold & Harms-Ringdahl, 1992). Prior to and immediately after the fatigue protocol, two 30s samples of quiet standing data were recorded.

At the end of the session, anthropometric measures were taken in order to build the 3D body model for the kinematics analysis. After adjusting the targets and positioning the subject, precise measurements of target coordinates and foot placements were taken and replicated in each subject's second session.

2.3. Data Acquisition

A TeleMyo sEMG measurement system (Noraxon USA Inc.) was used to record muscle activity. This system has an operating bandwidth of 10–350 Hz, an effective common mode rejection ratio of 130 dB DC, greater than 100 dB at 60 Hz, a minimum of 85 dB throughout the operating bandwidth, and a fixed overall per-channel gain of 2000. EMG data was sampled at 1080 Hz for four muscles of interest: upper trapezius (UT), anterior deltoid (AD), biceps (BIC) and triceps (TRIC), all collected from the reaching arm. A pair of electrodes was placed over each of the four muscles, with center-to-center distance of 3 cm oriented parallel to the muscle fibers. One electrode was placed as a reference over the C7 vertebra. Before placing the electrodes, the skin was cleaned with alcohol and shaved for better signal transmission. For the upper trapezius, the pair of electrodes were placed approximately 25 mm medial to the midpoint between C7 and angle of acromion. The location chosen for the anterior deltoid was 2 cm below the lateral third of the clavicle. For the biceps, the pair of electrodes were placed midway on the anterior part of the upper arm, over the muscle belly. For the triceps, the electrodes were placed 2 cm medial to the vertical midline of the posterior arm and midway between the acromion and the olecranon process (Basmajian & Blumenstein, 1980).

Six high-resolution cameras, part of the Vicon MX3 motion capture system (VICON©, Oxford Metrics ltd., Oxford, UK) were used to record the positions and movements of the whole body. A frequency of 120 Hz was used to sample kinematic data. To estimate the position and displacement of the CoM, a series of 49 passive and reflective markers were fixed to the skin using double-sided adhesive tape on different anatomical landmarks to recreate, in three dimensions, the segments of the human body. The body model is divided into 15 segments: (1) head (head to C7; 5 markers), (2-3) bilateral upper arms (shoulder to elbow; 3 markers per upper arm), (4-5) bilateral forearms (elbow to wrist; 4 markers per arm), (6-7) bilateral hands (distal to wrist; 4 markers per hand), (8) trunk (C7-T10; 8 markers), (9) pelvis (5 markers), (10-11) bilateral thighs (hip to knee; 3 markers per thigh), (12-

13) bilateral legs (knee to ankle; 3 markers) and (14-15) bilateral feet (3 markers per foot).

For kinetic recordings, two triaxial strain gauge force plates (AMTI© OR6-7, AMTI Inc, Watertown, USA) were used to measure the ground reaction forces and moments acting under the surface of each foot with a sampling frequency of 1080 Hz.

2.4. Data Analysis

For the RPT, the first trial of 30 s recorded was considered as No-fatigue (NF) and the last trial of 30 s as Fatigue-terminal (FT). These two trials, together with the two quiet standing trials recorded before and after the RPT were the four considered for further analyses.

Data from the NF and FT trials were separated in forward reaches (proximal to distal target) and backward reaches (distal to proximal target) using the target activation signals. EMG, kinematics and kinetics analysis for these two trials was performed on 10 complete forward reaches within these 30 s data blocks.

All analyzed EMG signals were first filtered using a dual-pass, 4th-order Butterworth filter, with a band-pass of 20–500 Hz. For the forward reaches analyzed, the root-mean-squared (RMS) amplitude of the four muscles were calculated and the average of the 10 forward reaches RMS was obtained. Matlab software was used to run the digital filter and to get the RMS values.

The kinematic variables of interest were identified by four markers, three segment angles and the body's CoM and were chosen based on previous studies showing significant fatigue effects at the completion of the RPT protocol on these variables (Emery & Côté, 2012; Emery, De Serres, McMillan, & Côté, 2010; Fuller, et al., 2009; Lomond & Côté, 2011). The four markers were right shoulder joint, left shoulder joint, right elbow joint and right index distal phalange. The three segment angles were the angles of the right scapula segment, the angles of the right shoulder and the angles of the right elbow. The coordinates of the four markers and of the CoM were obtained in the anterior-posterior

(AP), medial-lateral (ML) and superior-inferior (SI) directions, while the angles were obtained using Euler xyz rotations for scapula upward/downward rotation, anterior/posterior tilting and internal/external rotation (Emery, et al., 2010), shoulder flexion-extension and abduction-adduction, and elbow flexion-extension. Bodybuilder software (Vicon Motion Systems Ltd., Oxford, UK) was used to obtain these variables. Matlab software was used to run a digital dual-pass, 4th-order Butterworth filter with a frequency of 7 Hz and to obtain range of motion (ROM) (maximum – minimum position) and average position of the kinematic data over the duration of each considered forward pointing movement.

For kinetics, raw forces and moments were filtered using a dual-pass, 4th-order digital Butterworth filter with a frequency of 10 Hz. Kinetics data was obtained from the two fatigue trials and also from the two quiet standing trials in the anterior-posterior and medial-lateral direction. The outcome measures for the kinetics data (CoP) over the chosen forward movements were ROM, average position, displacement (RMS), maximum and minimum velocity, in each of mediolateral and anteroposterior directions, calculated over the considered forward reaches. Digital filtering and calculations were made with Matlab software (MathWorks, Massachusetts, USA).

2.5. Statistical Analysis

Descriptive statistics were computed (group average and SD measures). All the EMG, kinematic and kinetic parameters were analyzed using a two-way ANOVA with repeated measures having as independent factors the within-subject conditions of Belt (No-Belt and Belt) and Time (NF and FT). If a significant interaction effect of Belt by Time was found, Tukey's post-hoc comparisons were made to determine significant pair-wise differences ($p < 0.05$).

3. Results

3.1. Evidence of Fatigue

On average, subjects performed the repetitive pointing task for a time of 7.78 ± 4.69 min. For the No-Belt (NB) condition, the performance lasted an average time of 7.84 ± 4.42 min., while for the Belt (B) condition it was of 7.72 ± 5.06 min. Paired t-test analysis comparing the time to task termination in the NB condition versus the B condition showed no significant difference ($p = 0.85$). Also, there was no significant difference on average time to task termination between the first and second session ($p = 0.85$), which suggests a lack of learning effect or strategy from the subject that could affect the performance during the second session. For all the 19 participants the stoppage criteria met was a rate of perceived exertion equal or bigger than 8 on the Borg CR-10 scale.

Significant indicators of local muscle fatigue at the shoulder, neck and upper limb area were found. A main effect of Time was found for the anterior deltoid RMS ($p = 0.01$), biceps RMS ($p < 0.0005$) and upper trapezius RMS ($p < 0.05$). For the three muscles there was an increase of RMS in the last minute of the RPT (FT) compared to the first minute (NF). None of the four muscles analyzed showed a significant Belt effect.

3.2. Significant Belt x Time Interaction Effects

Statistical analysis of the kinetic data revealed a significant interaction effect in the maximum CoP velocity in the ML direction ($p < 0.04$) (Fig. 2). Post-hoc analysis showed that there was a decrease of the maximum CoP velocity in the NB condition at FT ($p = 0.03$), while for the B condition there was no significant difference in FT compared to NF ($p > 0.99$). It was also revealed that the maximum CoP velocity was bigger in the NB condition than in the B condition at both times (NF: $p < 0.0005$; FT: $p < 0.0005$).

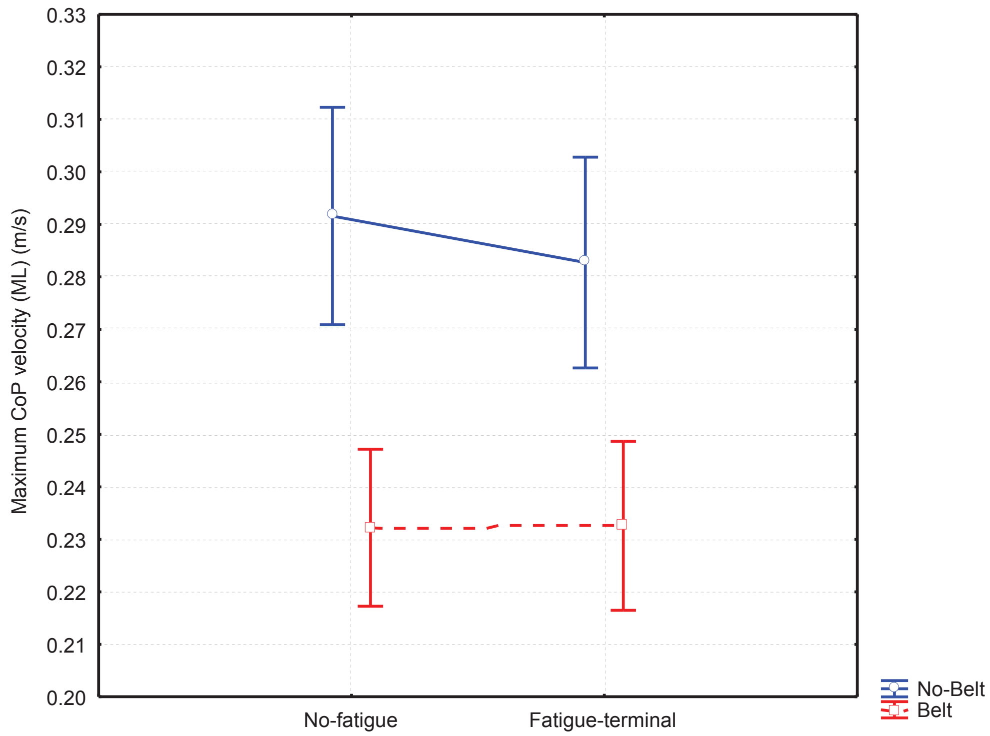

Fig. 2. Significant interaction effect in the maximum center of pressure (CoP) velocity in the medial-lateral (ML) direction for the No-Belt and Belt conditions during the first and the last minute of the repetitive pointing task.

An interaction effect was also found in the kinematic analysis. Figure 3 shows a significant interaction effect ($p = 0.01$) in the average reaching shoulder joint position in the SI direction. Post-hoc analysis revealed that the reaching shoulder was located significantly more superior with fatigue, in both the NB condition ($p < 0.0005$) and the B condition ($p < 0.0005$) and that this increased average position with fatigue was more pronounced in the NB condition (NB condition: increase of 10.6 ± 6.5 mm with fatigue; B condition: increase of 6.1 ± 4.5 mm with fatigue). Also, it was shown that the reaching shoulder joint was located significantly more inferior during the first ($p < 0.01$) and the last minute ($p < 0.0005$) of the fatigue protocol for the B condition in comparison to the NB condition.

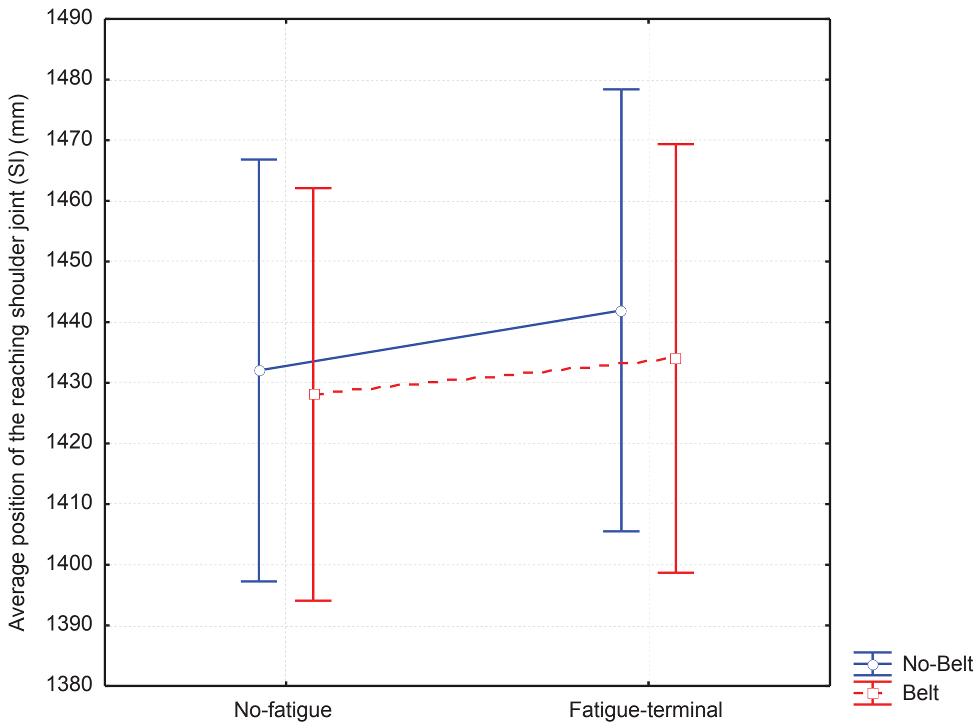

Fig. 3. Significant interaction effect in the average reaching shoulder joint position in the superior-inferior (SI) direction for the No-Belt and Belt conditions during the first and the last minute of the repetitive pointing task.

3.3 Significant Main Belt and Time Effects: Kinematic and Kinetic Average Positions

Table 1 summarizes the average positions of the different kinematic and kinetic parameters of interest displayed during the performance of the fatigue protocol.

Statistical analysis showed significant main effects of Time in the shoulder and elbow joints of the reaching arm as well as in the non-reaching shoulder joint. During fatigued movements, the reaching shoulder and elbow joints were displaced laterally towards the non-reaching side in the ML direction ($p < 0.0005$ and $p < 0.005$, respectively). Additionally, the reaching elbow joint was located on average more inferior in the SI direction ($p < 0.0005$). In the case of the average contralateral shoulder joint position, it was located less anterior in the AP direction ($p = 0.02$) when fatigue was induced.

A significant main effect of Time was also seen in different joint angles. In the presence of fatigue, subjects adopted a less upwardly rotated scapula average position ($p < 0.0001$) and a less abducted ($p < 0.0005$) and flexed ($p < 0.0001$) shoulder average position throughout the reaching movements.

Analyses also highlighted significant main effects of Belt. These were noted in the reaching shoulder and elbow joints, as well as in the CoM of the body. With the additional weight, there was a posterior shift of the average position of the reaching shoulder joint ($p < 0.03$), a lateral shift of the reaching elbow joint toward the non-reaching side ($p = 0.02$) and an inferior shift in the average CoM position ($p < 0.001$).

Table 1

Changes in kinematic average positions from No-fatigue (NF) to Fatigue-terminal (FT) in the No-Belt (NB) and Belt (B) conditions. The FT – NF mean differences (SD) are reported. Up/Down = Upward/Downward rotation; Ant/Post = Anterior/Posterior tilting; Int/Ext = Internal/External rotation; Abd/Add = Abduction/Adduction; Flex/Ext = Flexion/Extension; ns = not significant. Positive values indicate a more upwardly rotated, posterior tilted or external rotated scapula, a more abducted or flexed shoulder, a more flexed elbow, a more anterior, lateral (toward non-reaching arm), or superior position for the anterior-posterior (AP), medial-lateral (ML) and superior-inferior (SI) direction, respectively.

Parameter	Direction	NB (FT-NF)	B (FT-NF)	Belt *p*	Time *p*	Belt x Time *p*
Scapula segment angle (°)	Up/Down	-4.6 (2.6)	-3.1 (2.4)	ns	< 0.0005	ns
	Ant/Post	-0.5 (2.6)	0.6 (2.1)	ns	ns	ns
	Int/Ext	-1.3 (3.1)	-0.4 (3.4)	ns	ns	ns
Shoulder joint angle (°)	Abd/Add	-8.2 (6.6)	-6.5 (4.3)	ns	< 0.001	ns
	Flex/Ext	-6.6 (4.7)	-6.9 (4.1)	ns	< 0.0005	ns
Shoulder joint (mm)	AP	-7.7 (17.9)	-0.4 (13.0)	< 0.05	ns	ns
	ML	13.6 (10.7)	15.2 (13.9)	ns	< 0.001	ns
	SI	10.6 (6.5)	6.1 (4.5)	ns	ns	< 0.05
Elbow joint angle (°)	Flex/Ext	2.7 (14.1)	2.0 (4.5)	ns	ns	ns
Elbow joint (mm)	AP	-9.7 (16.2)	-4.9 (15.2)	ns	ns	ns
	ML	6.5 (9.5)	7.0 (11.2)	< 0.05	< 0.005	ns
	SI	-17.7 (12.8)	-16.0 (15.4)	ns	< .001	ns
Index distal phalange (mm)	AP	-1.3 (18.1)	-0.3 (14.5)	ns	ns	ns
	ML	-1.1 (11.6)	2.3 (6.5)	ns	ns	ns
	SI	-2.0 (9.5)	-1.8 (7.6)	ns	ns	ns
Shoulder joint (mm)	AP	-10.2 (15.9)	-10.0 (18.1)	ns	< 0.05	ns
(non-reaching arm)	ML	2.6 (11.2)	5.1 (10.8)	ns	ns	ns
	SI	-0.6 (11.0)	-3.4 (6.1)	ns	ns	ns
CoM (mm)	AP	-1.8 (12.4)	1.6 (7.0)	ns	ns	ns
	ML	2.6 (7.6)	5.2 (7.7)	ns	ns	ns
	SI	1.0 (2.9)	-0.3 (1.8)	< 0.001	ns	ns
CoP (mm)	AP	0.8 (6.2)	-0.8 (5.6)	ns	ns	ns
	ML	-0.2 (3.5)	-2.8 (14.2)	ns	ns	ns

3.4. *Significant Main Belt and Time Effects: Kinematic and Kinetic Ranges of Motion*

A summary of the ranges of motion (ROMs) of the kinematic and kinetic parameters of interest displayed during the fatigue protocol is showed in Table 2.

Significant effects of Time were found for ROM of all the kinematic parameters of interest except for the body's CoM. In the reaching arm and when fatigue was induced, increases of ROM were found in the shoulder and elbow joints in the AP direction ($p = 0.02$ and $p < 0.02$, respectively) and in the index distal phalange in the ML direction ($p < 0.01$). For the non-reaching shoulder joint, increases of ROM were also seen in the FT time in the AP and SI directions ($p = 0.01$ and $p < 0.0005$, respectively). Figure 4 illustrates an increased CoP ROM in the AP direction that was also found during the last minute of the fatigue protocol ($p < 0.01$). The ROM of the internal/external rotation angle of the reaching scapula increased with fatigue ($p < 0.005$), while the ROM of the reaching shoulder abduction/adduction angle ($p < 0.05$) and the ROM of the reaching elbow flexion/extension angle ($p < 0.005$) decreased.

Significant main effects of Belt were also revealed in the analysis. With the belt, the AP ROM of the non-reaching shoulder joint was lower than without the belt ($p < 0.05$) while the CoM ROM in the same direction was higher with the belt compared to without ($p < 0.04$).

Table 2

Changes in range of motion from No-fatigue (NF) to Fatigue-terminal (FT) in the No-Belt (NB) and Belt (B) conditions. The FT – NF mean differences (SD) are reported. Up/Down = Upward/Downward rotation; Ant/Post = Anterior/Posterior tilting; Int/Ext = Internal/External rotation; Abd/Add = Abduction/Adduction; Flex/Ext = Flexion/Extension; AP = anterior-posterior; ML = medial-lateral; SI = superior-inferior; ns = not significant.

Parameter	**Direction**	**NB (FT-NF)**	**B (FT-NF)**	**Belt *p***	**Time *p***	**Belt x Time *p***
Scapula segment angle (°)	Up/Down	-0.2 (1.3)	0.4 (1.3)	ns	ns	ns
	Ant/Post	0.4 (1.7)	0.6 (1.8)	ns	ns	ns
	Int/Ext	4.2 (5.3)	3.5 (4.2)	ns	< 0.005	ns
Shoulder joint angle (°)	Abd/Add	-1.3 (3.3)	-1.3 (2.8)	ns	< 0.05	ns
	Flex/Ext	-2.7 (9.1)	-1.5 (6.4)	ns	ns	ns
Shoulder joint (mm)	AP	8.4 (21.1)	10.6 (14.7)	ns	< 0.05	ns
	ML	1.1 (6.8)	3.3 (4.6)	ns	ns	ns
	SI	-0.7 (11.3)	1.7 (1.7)	ns	ns	ns
Elbow joint angle (°)	Flex/Ext	-4.8 (5.6)	-4.1 (4.9)	ns	< 0.005	ns
Elbow joint (mm)	AP	13.0 (23.7)	16.4 (24.6)	ns	< 0.05	ns
	ML	-2.9 (11.0)	-6.6 (12.1)	ns	ns	ns
	SI	-1.4 (16.1)	1.6 (8.7)	ns	ns	ns
Index distal phalange (mm)	AP	-1.5 (11.3)	0.3 (9.2)	ns	ns	ns
	ML	6.7 (12.4)	6.7 (8.2)	ns	< 0.01	ns
	SI	-1.3 (14.0)	2.3 (6.3)	ns	ns	ns
Shoulder joint (mm)	AP	10.1 (10.1)	4.3 (9.2)	< 0.05	< 0.05	ns
(non-reaching arm)	ML	2.3 (5.3)	4.0 (9.1)	ns	ns	ns
	SI	2.2 (2.1)	2.0 (2.6)	ns	< 0.001	ns
CoM (mm)	AP	-0.7 (5.0)	1.9 (4.7)	< 0.05	ns	ns
	ML	-0.6 (1.9)	1.3 (6.4)	ns	ns	ns
	SI	0.0 (1.6)	0.6 (0.7)	ns	ns	ns
CoP (mm)	AP	1.4 (2.8)	2.1 (3.7)	ns	< 0.01	ns
	ML	0.3 (1.2)	0.6 (1.6)	ns	ns	ns

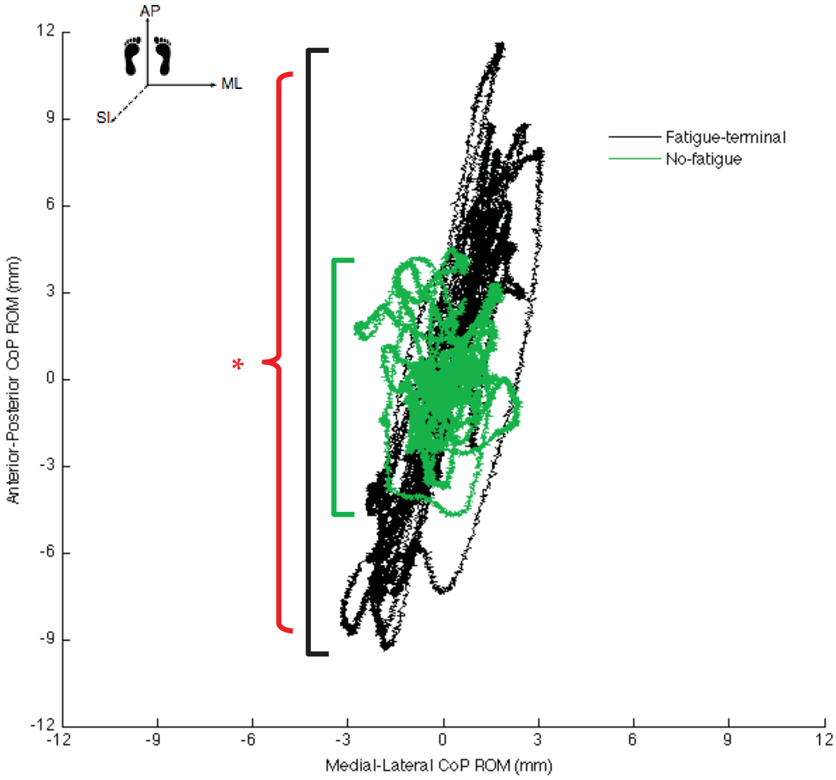

Fig. 4. Thirty-second sample of center of pressure range of motion (CoP ROM) for a representative subject (S9) in the Belt condition. CoP ROM increases significantly in the anterior-posterior direction in the last minute of the fatigue protocol, while there is no significant change in the medial-lateral direction.

4. Discussion

The purpose of this study was to determine the effects of adding weight to the trunk on the previously described whole-body adaptations to repetitive motion-induced arm fatigue accomplished from a standing position. Previous studies have shown that posture-movement adaptations occur in all planes of motion in an upper-limb task, suggesting voluntary strategies to contribute to reducing the load of the fatigued musculature while maintaining the goal of the upper limb movement task (Fuller, et al., 2009; Lomond & Côté, 2011). These studies have also shown increases in CoM and CoP movements, suggesting a mutual posture-movement assistance to adapt to fatigue. In the current study, we sought to determine whether challenging the postural component of the overall task would impact on these changes.

In the current study, the same repetitive upper limb task as that from previous studies was performed but the postural characteristics of the task were challenged by adding weight to the trunk. Thus, these two factors that affect postural control, fatigue and additional weight, were coupled in the same task, which to our knowledge, no study has investigated before. With the added weight, trunk movements are more destabilizing, so that it may not be as much a good strategy to jeopardize postural stability to have the trunk contribute to the arm task, as it had been observed in our previous studies (Fuller et al., 2009). Indeed, extra body weight has been identified as a factor that decreases postural stability (Hue, et al., 2007; Ledin, et al., 2004).

In our experiment, a belt was chosen to add weight because it is known that the center of mass of the human body while standing is located below the navel (McGinnis, 2005); thus, we took care to add weights in a configuration that would not change the location of the CoM. Previous studies have shown that a load of 15% extra body weight added to the trunk significantly changed postural angles (Ramprasad, et al., 2010) and postural control was affected with a larger body sway when the body weight was increased by 20% (Ledin, et al., 2004). Thus, a load of 20% was chosen for practical purposes in our study.

4.1. Evidence of Fatigue

The average time to fatigue and its standard deviation were similar to those observed in previous studies where the same upper-limb task was performed by groups of healthy young participants (7.9 ± 4.0 min) (Fuller, et al., 2009) (7.5 ± 3.0 min) (Lomond & Côté, 2011). No significant differences were seen in time to fatigue between the B and NB conditions, which was somehow expected since the load was not applied close to the muscles directly fatigued by the repetitive arm task. However, the fact that subjects were able to perform the task for the same duration even with the extra weight, theoretically representing a mechanically more challenging task, suggests an adaptability of the system, and will be discussed further below.

Our results show that EMG was affected by the repetitive task and that fatigue was successfully induced with the RPT protocol. The significant increases of EMG amplitude from the first to the last minute of the task for the anterior deltoid, biceps and upper trapezius are in line with results obtained in previous studies where the same task was performed (Fuller, et al., 2009; Lomond & Côté, 2011) and from studies where the task consisted of repetitive and sustained submaximal contractions (Fallentin, et al., 1993; Hagberg, 1981; Maton & Gamet, 1989; Mengshoel, Saugen, Forre, & Vollestad, 1995). Moreover, these results agree with the rates of perceived exertion (RPE) reported by the subjects according to the Borg CR-10 scale (Borg, 1982). Literature suggests the use of RPE as a reliable parameter for assessing the level of fatigue (Jones & Hunter, 1983).

4.2. Significant Belt x Time Interaction Effects

The maximum CoP velocity in the ML direction was the only parameter from the kinetic analysis that showed a significant Belt x Time interaction effect. Regardless of time, the maximum CoP velocity was lower when subjects carried the extra weight; however, fatigue only affected CoP velocity in the NB condition, reducing it. This happened even though neither fatigue nor the extra weight affected CoP ROM in that direction. A consequence of this could be that the decrease of CoP velocity caused by the extra load would induce a delay of the CoP following the CoM, which might increase the risk of injuries by falls. This could thus represent a mechanism by which extra weight increases risk of falls (Hue, et al., 2007), although further analyses are required to confirm this. Similarly, fatigue caused a decrease in the maximum CoP velocity in the ML direction for the NB condition, a result that is also related with the previous statement of increasing the risk of falling, since fatigue has also been identified as a potential risk factor for falls (Parijat & Lockhart, 2008). In the NB condition, subjects could have perceived that their postural stability was still well within a comfortable safety margin such that they could afford to reduce their CoP velocity as fatigue developed. However, when fatigued and having the extra weight, subjects may have perceived that further reducing CoP velocity could represent a significant threat to their postural stability and therefore avoided such a strategy and rather relied on other ones to adapt to fatigue. This interpretation is similar to the one made in previous studies using dual task paradigms, where postural adaptation strategies were observed to occur in directions less threatening to postural stability (Kang & Lipsitz, 2010; Vuillerme & Vincent, 2006). The current findings also support findings from these two studies in showing direction-specific adaptations to posture. Similarly, our results support our previously formulated hypothesis that the system is able to adapt its posture to upper limb fatigue but in a task-specific way (Fuller, et al., 2009).

Another significant interaction effect was found, with the average position of the shoulder joint of the reaching arm in the SI direction. At both time points, the shoulder joint of the participants' reaching arm was on average lower when they were wearing the belt. During the performance of the fatigue protocol and when carrying the extra load, participants may have slightly bent the body at the waist, with the upper back hunched caused by the extra weight. This could also explain the more inferior location of the reaching shoulder joint with the belt (main Belt effect) and may also explain the Belt main effect found in the average CoM position, which was also lower with additional weight. It is well known that any movement of the elemental units of the body (head, neck, trunk and limbs) in any direction results in a movement of the CoM in the same direction (McGinnis, 2005). Part of the reason why the CoM was located more inferior with the extra weight may be that the weights were added at a location slightly below the subject's non-weighted CoM, despite our best efforts to avoid this.

For both weight conditions the reaching shoulder was elevated with fatigue; however the elevation was more pronounced with fatigue in the absence of the additional weight. A similar strategy was seen in previous studies with healthy subjects performing the same repetitive task to fatigue, where the average shoulder joint position was moved upward and towards the non-reaching side (Fuller, et al., 2009;

Lomond & Côté, 2011). Authors explained this strategy as a compensation for the decrease in the average shoulder abduction angle and as a way to keep the upper limb away from the mesh barrier placed under the elbow, combining to reduce the load on the upper trapezius muscle by reducing the adduction moment produced by the mass of the upper arm about the shoulder joint (Fuller, et al., 2009). Moreover, as was the case for CoP velocity, the lack of extra weight showed a bigger effect of fatigue, and can also reflect a greater range of solutions to adapt to fatigue in the NB condition. However this did not represent a significant advantage to performance, since both belt conditions showed a similar time to fatigue.

Despite these explanations, the significant interaction effects indicate that fatigue had an effect on average shoulder height that was different when subjects were fitted with extra weight at the waist compared to when they were not. By elevating the shoulder with fatigue but less so with the added weight, subjects could have again perceived elevating the shoulder as either more threatening to postural stability or more mechanically difficult to achieve or sustain with the extra weight. This again suggests that few, but some differences in the adaptations to fatigue under the added weight condition were selected by the system, supporting the interpretation of task-specific adaptations to reach similar times to fatigue.

4.3. Significant Main Belt and Time Effects: Kinematic and Kinetic Average Positions

In addition to the effects discussed above, several other main effects of Time were found for average positions, which can be representative of postural adaptation strategies. The average position of the reaching elbow and shoulder was more posterior in FT. Authors suggest that this may correspond to a position where torques on the reaching shoulder are minimized, having as a consequence a more comfortable position for the participants when fatigued (Fuller, et al., 2009). Other postural changes in average positions of the reaching arm were observed in the frontal plane with fatigue. The reaching shoulder joint was located more toward the non-reaching side and subjects adopted a less abducted and flexed average shoulder angle that resulted in a more inferior location of the reaching elbow joint, although none of the subjects touched the barrier during the upper limb movement. The average position of the reaching elbow joint was also displaced laterally towards the non-reaching side. All of these adaptations are in line with a strategy of reducing the load on the upper trapezius, prolonging time to fatigue and keeping the reaching arm away from the mesh barrier under the elbow, as described previously (Fuller, et al., 2009; Lomond & Côté, 2011). These postural changes occurring in the frontal plane are in line with findings previously reported by Kang and Lipsitz (2010) where the system chose several postural adaptation strategies in the medial-lateral direction, the direction in which the stance base of support is the widest. However, compared to changes in average positions with fatigue described previously, we see fewer mediolateral postural adaptations in the present study. Indeed, Fuller et al. (2009) had observed lateral shifts toward the non-reaching side occurring with fatigue in several parameters, including in the non-reaching shoulder, CoM and CoP. This suggests that by adding data from the B condition, these Time effects disappear, again suggesting that with extra weight, subjects may move away from a strategy that could destabilize their posture. As alternate strategies, the main Belt effects on average positions show that the reaching shoulder is more posterior, the reaching elbow is more lateral or closer to the trunk, and the CoM is lower, which are all adaptations that tend to bring joints closer to an overall more stable body configuration.

4.4. Significant Main Belt and Time Effects: Kinematic and Kinetic Ranges of Motion

Many kinetic and kinematic parameters of interest showed an increase in ROM in the FT trials, which can represent movement strategies for other joints to further contribute to the pointing task in the presence of fatigue. In the AP direction, the reaching shoulder and elbow joints, the non-reaching shoulder joint and the CoP showed significant increases in ROM. These increases in joint linear ROM (excursion) were likely adopted to compensate for decreases in shoulder and elbow angular ROM seen with fatigue. Increases of ROM were also seen in the internal/external scapula rotation angle that may be explained by the natural movement of the task to be reaching back and forth that was more pronounced with fatigue as a strategy to facilitate the task with the help of the inertia obtained from the same spinning movement of the trunk. In comparison with results of our previous study (Fuller, et al., 2009), the same increases in ROM were observed to occur with fatigue in the current study except for the CoM, for which the presence of extra weight seems to have prevented increased ROM to occur. This suggests that the whole-body changes described in the present study may represent voluntary fatigue adaptations not only to contribute to the repetitive task as it is mentioned in previous studies (Côté, et al., 2008; Côté, et al., 2002; Fuller, et al., 2009), but also to maintain a stable and unchanged CoM in the presence of fatigue. This suggests that CoM could be considered as a global variable around which the system uses its redundancy in adapting various segments, with a goal of maintaining this global factor stable despite fatigue.

4.5. Limitations

Despite our efforts to maximize the motion capture system's potential (i.e. participants wearing tight clothes, daily calibration), a limitation of the study is the small displacements in the average positions and the small ranges of motion of the different kinematic parameters. While this may mean that the clinical meaningfulness of some of our results may be questioned, the fact that we did observe significant interaction, Belt and Time effects shows that these small displacements were constant in all or in the majority of the participants and therefore likely systematic. Another limitation is that we analyzed a limited number of kinematic parameters from the upper limbs and trunk, discarding joints that might be important for the analysis and interpretation of the results (i.e. wrist and lower limbs joints such as hips and ankles). Similarly, we chose to focus our verification of the fatigued status on the profiles of four muscles, such that the current analyses hardly allow us to make a parallel with multimuscle strategies that may underlie the observed kinematic changes. Finally, the limited number of motion capture cameras (6) might have represented a constraint to our accurate recording of the positions and movements in our whole-body model, which might have adversely affected our results.

5. Conclusion

Compared to the main fatigue effects, few significant main effects of Belt were found for kinetic and kinematic parameters in average position and range of motion. Most of them can likely be interpreted as direct mechanical consequence of wearing the belt, as discussed previously for the CoM. Taken together, the few main Belt and interaction effects seen in this study suggest that the fatigue adaptation strategies predominate over the threat that may pose the additional 20% body weight. With the few exceptions noted above, results suggest that despite changes of small amplitudes, fatigue adaptations described in this study are robust across subjects and postural conditions. However, it should be noted that for larger weight additions and/or tasks performed for a longer amount of time (e.g. carrying heavy loads or working with heavy protective clothing) the interactive effect of fatigue and extra weight might still constitute a risk factor for falls or injuries (Hue, et al., 2007; Ledin, et al., 2004; Parijat & Lockhart, 2008; Tjepkema, 2003). The CoM range of motion increases with fatigue seen in a previous study but not here suggests that with additional weight, the system may develop fatigue adaptation strategies that avoid disturbing this postural characteristic. More studies are needed, perhaps with higher postural threats, to better understand the interaction between fatigue and body weight, so as to better predict and avoid injuries by falls.

Acknowledgements

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CONCLUSION

The principal objective of this thesis was to quantify the effects of additional weight applied to the trunk on posture and movement adaptations to repetitive arm motion-induced fatigue. We had previously shown that lateral postural shifts and increased postural range of motion occurred as fatigue developed while healthy subjects accomplished a repetitive pointing task from a standing position. We therefore sought to verify if the same fatigue adaptation strategies would be developed in a situation where such postural adaptations would represent an extra threat to equilibrium, in a case where extra weight was added to the trunk.

As a result of the repetitive pointing task (RPT), signs of fatigue were exhibited, including increases in ratings of perceived exertion and in shoulder muscle activity. In addition, a few Belt by Time interaction effects were found. The maximum center of pressure (CoP) velocity was lower with the added weight at both time points (No-fatigue and Fatigue-terminal); however, with fatigue, it stayed constant in the added weight condition (Belt), while it decreased in normal conditions (No-Belt). The reaching shoulder's average vertical position was lower in the Belt condition, regardless of time, and showed a higher increase with fatigue in the No-Belt condition. Similarly to our previous studies, medial-lateral shifts in the average positions of the reaching shoulder and elbow joints were seen, but they were not seen in the center of mass (CoM) and CoP. CoM range of motion (ROM) did not change significantly while the ROM of all the other parameters did.

Findings of this research support the belief that the system possesses a large repertoire of strategies to compensate for challenging conditions such as fatigue and increased postural threat by developing some common but also some different solutions to reach the same performance (in our case, similar endurance times). More specifically to the current study, findings suggest that fatigue adaptation strategies predominate over postural stabilization strategies, even when postural stability is further challenged, as in this case, by additional weight. As was suggested by our previous studies, we believe that posture-movement adaptations are used as a strategy to reduce the load on the fatigued musculature, while at the same time maintaining the body safe from a postural threat. However, with a greater threat to postural stability (i.e. higher additional weight or applying a postural perturbation), other findings could emerge. Fatigue and additional weight both represent threat to postural stability. More studies are needed to gain a better understanding of how they interact, so that postural instability related injuries could be prevented.

The results of this research could provide important knowledge for the prevention of injuries and for the maintenance of a safe working posture in workplaces where persons are required to perform a repetitive task while carrying a load weight (i.e. firefighter, army) or in unstable environments (train, boat, airplane workers). Our results apply for conditions where persons accomplish their work in a standing position; to understand adaptations to fatigue and additional weight in a different stance condition (i.e. seated or during gait), more work needs to be done. In the same way, this research could have important implications in the treatment and management of diseases linked to postural instability and fatigue, such as in people with chronic fatigue or people with obesity. Therefore, results of such studies could be used by ergonomists and occupational therapists in designing interventions that could help prevent risk of falls related to fatigue in such individuals.

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APPENDICES

A. Ethics Certificate

Comité d'éthique de la recherche
des établissements du CRIR

CRIR

Certificat d'éthique

Par la présente, le comité d'éthique de la recherche des établissements du CRIR (CÉR) atteste qu'il a évalué, par voie accélérée, le projet de recherche **CRIR-504-0410** intitulé :

« Experimental Analysis of Whole-body Coordination Changes Associated with Repetitive Upper Limb Motion : What, When, Why ?».

Présenté par : **Julie Côté**

Le présent projet répond aux exigences éthiques de notre CÉR. Le Comité autorise donc sa mise en œuvre sur la foi des documents suivants :

- Formulaire A daté du 11 avril 2010 ;
- Protocole de recherche intitulé « Experimental Analysis of Whole-body Coordination Changes Associated with Repetitive Upper Limb Motion : What, When, Why ?» ;
- Preuve d'octroi de fonds- NSERC Discovery Grants 2010 results (March 25th 2010) ;
- Lettre de l'Hôpital juif de réadaptation, datée du 29 avril 2010, attestant que l'établissement accueille favorablement le projet sur le plan de la convenance institutionnelle ;
- Formulaire de consentement (version approuvée du 22 avril 2010) ;
- Consent form (version approuvée du 22 avril 2010) ;
- Affiche de recrutement- version française du 22 avril 2010 ;
- Affiche de recrutement- version anglaise du 22 avril 2010.

Ce projet se déroulera dans le site du CRIR suivant : **Hôpital juif de réadaptation.**

Ce certificat est valable pour un an. En acceptant le présent certificat d'éthique, le chercheur s'engage à :

1. Informer, dès que possible, le CÉR de tout changement qui pourrait être apporté à la présente recherche ou aux documents qui en découlent (Formulaire M) ;
2. Notifier, dès que possible, le CÉR de tout incident ou accident lié à la procédure du projet ;
3. Notifier, dès que possible, le CÉR de tout nouveau renseignement susceptible d'affecter l'intégrité ou l'éthicité du projet de recherche, ou encore, d'influer sur la décision d'un sujet de recherche quant à sa participation au projet ;
4. Notifier, dès que possible, le CÉR de toute suspension ou annulation d'autorisation relative au projet qu'aura formulée un organisme de subvention ou de réglementation ;
5. Notifier, dès que possible, le CÉR de tout problème constaté par un tiers au cours d'une activité de surveillance ou de vérification, interne ou externe, qui est susceptible de remettre en question l'intégrité ou l'éthicité du projet ainsi que la décision du CÉR ;
6. Notifier, dès que possible, le CÉR de l'interruption prématurée, temporaire ou définitive du projet. Cette modification doit être accompagnée d'un rapport faisant état des motifs à la base de cette interruption et des répercussions sur celles-ci sur les sujets de recherche ;

7. Fournir annuellement au CÉR un rapport d'étape l'informant de l'avancement des travaux de recherche (formulaire R) ;
8. Demander le renouvellement annuel de son certificat d'éthique ;
9. Tenir et conserver, selon la procédure prévue dans la *Politique portant sur la conservation d'une liste des sujets de recherche*, incluse dans le cadre réglementaire des établissements du CRIR, une liste des personnes qui ont accepté de prendre part à la présente étude ;
10. Envoyer au CÉR une copie de son rapport de fin de projet / publication.

Me Michel T. Giroux
Président du CÉR

Date d'émission
29 avril 2010

B. Consent Form (English version)

Centre de recherche interdisciplinaire en réadaptation du Montréal métropolitain

Consent form

Hôpital juif de réadaptation
Jewish Rehabilitation Hospital

1 - Title of project

Experimental analysis of whole-body coordination changes associated with repetitive upper limb motion: what, when, why? (CRSNG-RGPIN 312333-05)

2 - Researchers in charge of project

Julie Côté, Ph.D., Associate Professor, Department of Kinesiology and Physical Education, McGill University, (514) 398-4184 ext. 0539, (450) 688-9550, ext. 4813

Kim Emery, M.Sc., research assistant, Jewish Rehabilitation Hospital research center, (450) 688-9550 ext. 4827

3 - Preamble/Introduction

We are asking you to participate in a research project involving the analysis of your movements when you are fatigued. Before agreeing to participate in this project, please take the time to study and carefully consider the following information.

This consent form explains the aim of this study, the procedures, advantages, risks and drawbacks, as well as the persons to contact, if necessary.

This consent form may contain words that you do not understand. We invite you ask any question that you deem useful to the researcher and the others members of the staff assigned to the research project and ask them to explain any word or information which is not clear to you.

4 - Project description and objectives

The objectives of this research are to describe the time sequence of within- to between muscle changes occurring with repetitive motion-induced fatigue, and to assess the effects of fatigue on postural and proprioceptive characteristics. This project also aims at assessing gender differences in these fatigue responses. 24 healthy subjects will be recruited for this project (12 men, 12 women) and will perform a laboratory assessment protocol twice, with at least 48 hours in between. The long-term objective of this project is to better understand how humans coordinate posture and repetitive arm motion and how fatigue affects this coordination. This research will provide knowledge and tools to identify, treat and prevent musculoskeletal disorders.

5 - Nature and duration of participation

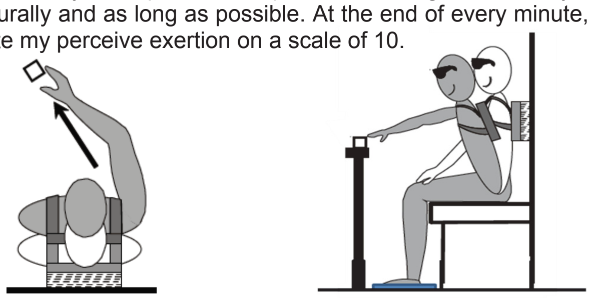
The research project to which I am invited to participate aims at understanding how we coordinate posture and repetitive arm motion and how fatigue may affect this coordination. The experimental procedure takes place at the research center of the Jewish Rehabilitation Hospital. I am asked to participate in two experimental sessions, 48 hours apart, which will last approximately two hours each. Each of the two experimental sessions will be separated in four different phases:

Phase 1: preparation (30 minutes),
Phase 2: pre-fatigue tests (30 minutes),
Phase 3: fatigue procedure (20 minutes) and
Phase 4: post-fatigue tests (30 minutes).

During Phase 1, surface electrodes will be applied on the skin over my neck and dominant upper limb muscles in order to measure their activity. Reflective markers will be fixed on the skin over my neck, trunk, arms and legs in order to record their positions. A heart rate monitor will be placed on my chest with an elastic band. None of these procedures is invasive.

During Phase 2, I will be asked to push upward with my dominant shoulder as hard as I can, against a rigid frame. I will be asked to repeat the task but this time, pushing at 30% of my maximum. Then, I will be asked to perform 10 consecutive reaches with my dominant arm moving between two targets, following the beat of a metronome, and then at the same rhythm but without the metronome, eyes closed. Then, I will be asked to stand as stable and symmetrically as possible. Finally, I will be asked to rest my dominant arm on a movable table. The table will move and I will have to push a button when I will feel that my arm has reached a horizontal position. For each task, I will perform three consecutive trials, with rest in between.

During Phase 3, I will stand on two force plates placed on the floor and I will wear a harness. The harness will restrict (session 1) or not (session 2) the movements of my trunk (see Figure below; however for this study, I will be standing, not seated). These two sessions will be assigned in a random order. Once I will be ready, I will perform a repetitive reaching task with my dominant arm, as naturally and as long as possible. At the end of every minute, I will be asked to rate my perceive exertion on a scale of 10.



During Phase 4, I will perform the same tests as those performed in Phase 1. Between trials, I will keep my dominant arm horizontal, at shoulder height. This experimental procedure will be repeated in identical fashion (except for the trunk harness in the fatigue protocol) during the 2^{nd} experimental session conducted a minimum of 48 hours later.

6 - Advantages associated with my participation

I will not personally benefit from any advantage by participating in this study. However, I will contribute to the advancement of knowledge of human movement and musculoskeletal disorders.

7 - Risks associated with my participation

None of the techniques used are invasive. I understand that my participation in this project does not put me at any medical risk.

8 - Personal inconvenience

The duration of each experimental session (approximately 2 hour each) may represent an inconvenience for me. The possibility that some small regions (8, 3x3 cm each) of the skin over my neck and arm muscles have to be shaven before placing the electrodes might also represent an inconvenience for me. Although it is hypo-allergenic, the adhesive tape used to fix the electrodes on my skin may occasionally produce some slight skin irritation. Should this happen, a hypo-allergic lotion will be applied on my skin to relieve skin irritation. Also, I will experience some fatigue towards the end of each protocol, which may cause some tenderness, stiffness and/or pain in the neck-shoulder area during and/or following the session. These symptoms should dissipate within 48 hours following the completion of the protocol. A clinician will be present at all times during the experimental sessions.

9 - Access to my medical file

No access to my medical file is required for this study.

10 - Confidentiality

All the personal information collected for this study will be codified to insure confidentiality. Information will be kept under locking key at the research center of the Jewish Rehabilitation Hospital by one of the persons responsible for the study for a period of five years following the end of the study. Only the people involved in the project will have access to this information. If the results of this research project are presented or published, nothing will allow my identification. After this five-year period, data will be destroyed.

11 - Questions concerning the study

The researchers present during the testing should answer my questions concerning the project in a satisfactory manner.

12 - Withdrawal of subject from study

Participation in the research project described above is completely voluntary. I have the right to withdraw from the study at any moment. If ever I withdraw from the study, all documents concerning myself will be destroyed.

13 - Responsibility

By accepting to enter this study, I do not surrender to my rights and do not free the researchers, sponsor or the institutions involved from their legal and professional obligations.

14 - Monetary compensation

No monetary compensation will be given to me for participation in this protocol.

15 - Contact persons

If I need to ask questions about the project, signal an adverse effect and/or an incident, I can contact at any time Julie Côté, Ph.D., or Kim Emery, M.Sc., at the numbers indicated on the 1st page. I may also contact M. Michael Greenberg, local commissioner for the quality of services at the JRH, at (450) 688-9550, extension 232.

Also, if I have any questions concerning my rights regarding my participation to this research project, I can contact Mme. Anik Nolet, Research ethics co-ordinator of CRIR at (514) 527-4527 ext. 2643 or by email at anolet.crir@ssss.gouv.qc.ca

Experimental analysis of whole-body coordination changes associated with repetitive upper limb motion: what, when, why? (CRSNG-RGPIN 312333-05)

CONSENT

I declare to have read and understood the project, the nature and the extent of the project, as well as the risks and inconveniences I am exposed to as described in the present document. I had the opportunity to ask all my questions concerning the different aspects of the study and to receive explanations to my satisfaction.

I, undersigned, voluntarily accept to participate in this study. I can withdraw at any time without any prejudice. I certify that I have received enough time to take my decision.

A signed copy of this information and consent form will be given to me.

NAME OF PARTICIPANT (print): ______________________________

SIGNATURE OF PARTICIPANT: ______________________________

SIGNED IN ____________________, on ________________, 20_____.

COMMITMENT OF RESEARCHER

I, undersigned, ______________________________ , certify

(a) having explained to the signatory the terms of the present form ;

(b) having answered all questions he/she asked concerning the study ;

(c) having clearly told him/her that he/she is at any moment free to withdraw from the research project described above; and

(d) that I will give him/her a signed and dated copy of the present document.

____

Signature of person in charge of the project or representative

SIGNED IN _________________, on ________________________ 20__.

Research protocol approved by the Committee for research ethics of the CRIR establishments, on xx/xx/2010

CRIR Centre de recherche interdisciplinaire en réadaptation du Montréal métropolitain

Formulaire de consentement

Hôpital juif de réadaptation Jewish Rehabilitation Hospital

1 - Titre du projet

Analyse expérimentale des changements de coordination globale associés au mouvement répétitif du membre supérieur : quoi, quand, pourquoi ? (CRSNG-RGPIN 312333-05)

2 - Responsable(s) du projet

Julie Côté, Ph.D., professeure agrégée, département de kinésiologie et d'éducation physique, université McGill, (514) 398-4184 poste 0539, (450) 688-9550, poste 4813

Kim Emery, M.Sc., assistante de recherche, centre de recherche de l'hôpital juif de réadaptation, (450) 688-9550 poste 4827

3 - Préambule/Introduction

Nous vous demandons de participer à un projet de recherche qui implique l'analyse de vos mouvements lorsque vous êtes fatigué. Avant d'accepter de participer à ce projet de recherche, veuillez prendre le temps de comprendre et de considérer attentivement les renseignements qui suivent.

Ce formulaire de consentement vous explique le but de cette étude, les procédures, les avantages, les risques et inconvénients, de même que les personnes avec qui communiquer au besoin.

Le présent formulaire de consentement peut contenir des mots que vous ne comprenez pas. Nous vous invitons à poser toutes les questions que vous jugerez utiles au chercheur et aux autres membres du personnel affecté au projet de recherche et à leur demander de vous expliquer tout mot ou renseignement qui n'est pas clair

4 - Description du projet et de ses objectifs

Les objectifs de cette recherche sont de décrire la séquence des changements intra- et inter-musculaires, ainsi que de mesurer les changements posturaux et proprioceptifs occasionnés par la fatigue due au mouvement répétitif. De plus, le projet vise à évaluer les différences de réponse à la fatigue entre les hommes et les femmes. 24 sujets en santé seront recrutés (12 hommes et 12 femmes) et participeront à un protocole d'évaluation en laboratoire deux fois, espacées d'au moins 48 heures.

L'objectif à long terme de cette recherche est de mieux comprendre comment l'humain coordonne sa posture avec les mouvements répétitifs du bras, et comment la fatigue influence cette coordination. Cette étude permettra aussi d'accroître les connaissances et d'identifier des outils pour la prévention et le traitement des blessures musculo-squelettiques.

5 - Nature et durée de la participation

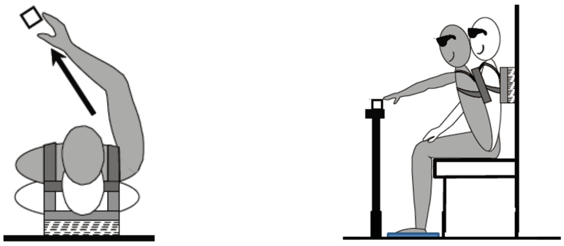
Le projet de recherche auquel je suis invité(e) à participer vise à comprendre comment nous coordonnons notre posture et nos mouvements répétitifs du bras et comment la fatigue peut influencer cette coordination. La procédure expérimentale sera effectuée au centre de recherche de l'hôpital juif de réadaptation. On me demande de participer à deux séances expérimentales, espacées d'au moins 48 heures, d'une durée approximative de 2 heures chacune. Chaque séance comportera quatre phases :

Phase 1 : préparation (30 minutes),
Phase 2 : tests pré-fatigue (30 minutes),
Phase 3 : procédure de fatigue (20 minutes) et
Phase 4 : tests post-fatigue (30 minutes).

Durant la Phase 1, des électrodes seront fixées sur la peau des muscles de ma colonne et de mon bras dominant afin de mesurer leur activité. Des marqueurs réfléchissants seront fixés sur la peau de mon cou, de ma colonne, de mes bras et de mes jambes afin d'enregistrer leurs déplacements. Un cardiofréquencemètre sera installé sur ma poitrine à l'aide d'une bande élastique. Aucune de ces procédures n'est effractive.

Lors de la Phase 2, on me demandera de pousser vers le haut avec mon épaule dominante, le plus fort possible, contre un appareil fixe. Ensuite, je devrai refaire le test en poussant à 30% de ma force maximale. Ensuite, on me demandera de faire 10 mouvements consécutifs, mon bras dominant se déplaçant entre deux cibles, suivant le rythme d'un métronome, et ensuite sans le métronome, les yeux fermés. Ensuite, on me demandera de me tenir debout sur deux plateformes de force, le plus stable et symétriquement possible. Finalement, on me demandera de placer mon bras sur une table amovible. Cette table se déplacera et je devrai appuyer sur bouton poussoir le moment où je perçois que mon bras est en position horizontale. Pour chaque tâche, j'effectuerai trois essais, avec du repos entre chaque.

Lors de la Phase 3, je serai debout sur deux plateformes de force placées sur le sol et je porterai un harnais. Le harnais restreindra (séance 1) ou non (séance 2), les mouvements de mon tronc (voir Figure, page suivante; cependant pour cette étude, je serai debout, et non assis). L'ordre des séances sera établi au hasard. Une fois installé, on me demandera d'effectuer un mouvement répétitif à la hauteur de l'épaule avec mon bras dominant, le plus naturellement et longtemps possible. À la fin de chaque minute, je devrai identifier mon niveau d'effort perçu, sur une échelle de 0 à 10.



Lors de la Phase 4, on me demandera de refaire les tests effectués lors de la Phase 1. Entre les essais, je maintiendrai mon bras dominant à l'horizontale, à la hauteur de l'épaule. Cette procédure expérimentale sera répétée de façon identique (excepté pour l'utilisation du harnais) lors d'une deuxième séance qui aura lieu un minimum de 48 heures plus tard.

6 - Avantages pouvant découler de ma participation

Je ne retirerai personnellement aucun avantage à participer à cette étude. Toutefois, j'aurai contribué à l'avancement des connaissances portant sur le mouvement humain et les blessures musculo-squelettiques.

7 - Risques pouvant découler de ma participation

Aucune des procédures décrites n'est effractive. Je comprends que ma participation à cette recherche ne me fait courir aucun risque médical.

8 - Inconvénients personnels

La durée de chaque séance expérimentale (environ deux heures chacune) peut représenter un inconvénient pour certaines personnes. La possibilité que quelques petites surfaces (8, 3x3 cm each) de la peau sur les muscles de mon cou et de mon bras doivent être rasées avant d'y apposer les électrodes peut aussi représenter un inconvénient pour moi. Bien qu'il soit hypo-allergène, le ruban adhésif utilisé pour maintenir les électrodes sur la peau peut occasionnellement provoquer de légères irritations de la peau. Le cas échéant, une lotion hypo-allergène sera appliquée pour soulager l'irritation cutanée. Aussi, je vais ressentir de la fatigue vers la fin de la séance expérimentale, ce qui pourrait occasionner de la sensibilité, de la raideur et/ou de la douleur dans la région du cou et de l'épaule durant et/ou après la séance. S'ils se manifestent, ces symptômes devraient disparaître dans les 48 heures suivant la fin du protocole expérimental. Un clinicien sera présent en tout temps pendant les séances expérimentales en cas de complications.

9 - Accès à mon dossier médical

Aucun accès à mon dossier médical n'est requis pour cette étude.

10 - Confidentialité

Tous les renseignements personnels recueillis à mon sujet au cours de l'étude seront codifiés afin d'assurer la confidentialité. Ces données seront conservées au centre de recherche de l'Hôpital juif de réadaptation et gardées sous clé par un responsable de l'étude pour une période de cinq ans suivant la fin de l'étude. Après cette période, les données seront détruites. Seuls les membres de l'équipe de recherche y auront accès. En cas de présentation des résultats de cette recherche sous forme écrite ou orale, rien ne pourra permettre de m'identifier.

11 - Questions concernant cette étude

Les chercheurs présents lors de la collecte des données s'engagent à répondre de façon satisfaisante à toutes mes questions concernant le projet de recherche.

12 - Retrait de la participation du sujet

Ma participation au projet de recherche décrit ci-dessus est tout à fait libre et volontaire. Il est entendu que je pourrai, à tout moment, mettre un terme à ma participation. En cas de retrait de ma part, les documents électroniques et écrits me concernant seront détruits.

13 - Clause de responsabilité

En acceptant de participer à cette étude, je ne renonce à aucun de mes droits ni ne libère les chercheurs, le commanditaire ou les institutions impliquées de leurs obligations légales et professionnelles.

14 - Indemnité compensatoire

Aucune compensation financière ne me sera offerte pour ma participation à cette étude.

15 - Personnes ressources

Si je désire poser des questions sur le projet ou signaler des effets secondaires, je peux rejoindre en tout temps Julie Côté, Ph.D. ou Kim Emery, M.Sc., aux numéros indiqués à la 1ère page. Je peux également contacter Monsieur Michael Greenberg, commissaire locale à la qualité des services de l'HJR, au (450) 688-9550 poste 232.

De plus, si j'ai des questions sur mes droits et recours ou sur ma participation à ce projet de recherche, je peux communiquer avec Me Anik Nolet, coordonnatrice à l'éthique de la recherche des établissements du CRIR au (514) 527-4527 poste 2643 ou par courriel à l'adresse suivante: anolet.crir@ssss.gouv.qc.ca

CONSENTEMENT

Je déclare avoir lu et compris le présent projet, la nature et l'ampleur de ma participation, ainsi que les risques auxquels je m'expose tels que présentés dans le présent formulaire. J'ai eu l'occasion de poser toutes les questions concernant les différents aspects de l'étude et de recevoir des réponses à ma satisfaction.

Je, soussigné(e), accepte volontairement de participer à cette étude. Je peux me retirer en tout temps sans préjudice d'aucune sorte. Je certifie qu'on m'a laissé le temps voulu pour prendre ma décision.

Une copie signée de ce formulaire d'information et de consentement doit m'être remise.

NOM DU SUJET ____

SIGNATURE ____

Signé à ________________________, le ___________, 20_____.

ENGAGEMENT DU CHERCHEUR

Je, soussigné (e), ________________________________ , certifie

(a) avoir expliqué au signataire les termes du présent formulaire;

(b) avoir répondu aux questions qu'il m'a posées à cet égard;

(c) lui avoir clairement indiqué qu'il reste, à tout moment, libre de mettre un terme à sa participation au projet de recherche décrit ci-dessus;

et (d) que je lui remettrai une copie signée et datée du présent formulaire.

____

Signature du responsable du projet ou de son représentant

Signé à _________________, le ______________ 20__.