USE OF ARMS – INFLUENCE TO HUMAN BALANCE DURING PERTURBATION

Kirsi Yrjölä

Master's thesis
Biomechanics
Autumn 2014
Department of Biology of Physical Activity
University of Jyväskylä
Supervisors:
Vesa Linnamo
Andrew Cresswell
Jarmo Piirainen

Kirsi Yrjölä (2014). Use of arms – influence to the human balance during perturbation. Department of Biology of Physical Activity, University of Jyväskylä, Master's Thesis in Biomechanics, 56 p.

ABSTRACT

The purpose of this study was to investigate how the use of arms influences to the human balance control while random perturbation.

Five voluntary subjects (32 ± 11 years) participated to this study. For the balance control during perturbation the subjects had to stand still on a treadmill surface which was moving anterior (velocity 0.24 m/s) direction by 0.2 m providing the unbalanced situation. Study was performed in Arms (normal stance) and No Arms –condition where the subject crossed his / her arms in front of the body. The balance was measured by a force plate under the treadmill and subjects center of pressure (COP) was also used for a trigger for the EMG analyses, which was calculated as RMS amplitude with 300 ms window from Soleus (SOL), Medialis Gastrocnemius (MG), Tibialis Anterior (TA), Rectus Abdominis (RA), Erector Spinae (ES), Deltoideus Anterior (AD) and Deltoideus Posterior (PD) muscles. The EMG amplitude data was also taken into further analyze by diving it into three time windows after onset (0 to 50 ms = EMG_{50}, 50 to 150 ms = EMG_{150} and 150 to 300 ms = EMG_{300}, respectively). Kinematics were recorded continuously and COP was controlled by an instrumented force sensing treadmill and custom written software.

The main finding from this study is that there is no significant differences in balance control weather you use or not use your arms while perturbation occurs anteriorly. Nevertheless, a higher EMG activity of the upper body muscles were observed when not using arms (No Arms –condition) compared to the normal stance with free use of arms. The EMG of RA and AD muscles had a significant role in the balance control -system when not using arms ($p < 0.05$). Timeslot analyze suggested that the RA EMG_{150} was 59 ± 2 % higher ($p < 0.05$) when no arms were used and AD EMG increased by 22 ± 19 % ($p < 0.05$) from the EMG_{50} to the EMG_{300} in No Arms -condition. The EMG of the ES muscle was on average two times higher in No Arms –condition but the individual variations were remarkable. Based on the results the EMG activity of the upper body muscles is higher when not using arms compared to the situation where arms are normally in use. Because of the higher activation level needed from the upper body muscles during perturbation, one should take care of the adequate muscle strength.

Key words: Balance, Upper body, 3D, COP, EMG

Kirsi Yrjölä (2014). Käsien vaikutus ihmisen tasapainoon häiriötilanteessa. Liikuntabiologian laitos, Jyväskylän yliopisto. Biomekaniikan Pro gradu –tutkielma, 56s.

TIIVISTELMÄ

Tämän Pro gradu –tutkielman tarkoituksena oli selvittää, miten käsien käyttäminen tai käyttämättömyys vaikuttaa ihmisen tasapainon korjaamiseen äkillisessä horjutus (häiriö) tilanteessa.

Tutkimukseen osallistui viisi vapaaehtoista koehenkilöä (32 ± 11 vuotta). Tutkimuksen testiprotokollassa tasapainon korjaamiseksi häiriötilanteessa, henkilö asetettiin seisoman juoksumaton päälle, joka liikahti 0.2 metriä eteenpäin, anterior-suunnassa (nopeus 0.24 m/s). Tutkimuksessa käytettiin kahta eri tilaa: kädet ja ei kädet –tila, missä koehenkilö sulki kätensä vartalon eteen puuhkaan. Tasapainoa mitattiin juoksumaton voimalevyltä ja koehenkilön painopisteen muutokset (COP) analysoitiin. Painopistettä (COP) käytettiin myös triggerinä EMG analyysia varten, joka toteutettiin 300 millisekunnin aikaikkunoilla Soleus (SOL), Medial Gastrocnemius (MG), Tibialis Anterior (TA), Rectus abdominis (RA), Erector Spinae (ES), Anterior Deltoid (AD) ja Posterior Deltoid (PD) lihaksista. Lopputuloksissa EMG data jaettiin vielä kolmeen eri aikaikkunaan (0 – 50 ms = EMG_{50}, 50 – 150 ms = EMG_{150} sekä 150 – 300 ms = EMG_{300}) tarkempaa analyysiä varten. Kinematiikkadata tallennettiin jatkuvana. Koehenkilön painopistettä (COP) kontrolloi voimalevyillä varustettu juoksumatto sekä tähän tutkimukseen räätälöity erikoissofta.

Tutkimuksen päätuloksena voi esittää sen, että käsien käyttämisellä ei ole tilastollisesti merkittävää vaikutusta tasapainon korjaamiseen häiriötilanteessa anterior-suunnassa. Tuloksissa on kuitenkin nähtävissä merkittävästi suuremmat EMG aktivaatiotasot ylävartalon lihaksissa ($p < 0.05$) silloin, kun käsiä ei käytetä (ei kädet –tila) verrattuna kädet-tilaan. Yläkropan RA ja AD lihasten aktivaatiotasot vaikuttivat merkitsevästi tasapainon kontrollointiin tilassa, missä käsiä ei käytetty ($p < 0.05$). Kun tuloksia pilkottiin ja analysoitiin pienemmissä ajanjaksoissa, oli RA lihaksen EMG_{150} 59 ± 2 % suurempi ($p < 0.05$) tilassa, missä käsiä ei käytetty ja AD lihaksen EMG kasvoi 22 ± 19 % ($p < 0.05$), kun verrattiin ajanjaksoa EMG_{50} ajanjaksoon EMG_{300}, tilassa missä käsiä ei käytetty. ES lihaksesta mitattu EMG aktivaatio oli keskimäärin noin kaksinkertaista ei kädet –tilassa, mutta yksilölliset vaihtelut olivat huomattavan suuria. Tutkimuksen tulosten perusteella voidaan todeta, että ihmisen tasapainon kontrolloinnissa, tilassa missä kädet eivät ole käytössä, ovat ylävartalon lihasten aktivaatiotasot suurempia verrattuna normaaliin kädet käytössä tilaan. Koska ylävartalon lihasaktivaatio on suurempaa häiriötilassa ilman käsiä, tulisi huolehtia riittävistä lihasten voimantuotto-ominaisuuksista tasapainon korjaamisessa.

Avainsanat: Balance, Upper body, 3D, COP, EMG

ABBREVIATIONS

AD	Anterior deltoid
AP	Anterior posterior
BOS	Base of support
CNS	Central nervous system
COM	Center of mass
COP	Center of pressure
EMG	Electromyography
ES	Erector spinae
MG	Medial gastrocnemius
ML	Medialis lateralis
MVC	Maximal voluntary contraction
PD	Posterior deltoid
RA	Rectus abdominis
RMS	Root mean square
SOL	Soleus
TA	Tibialis anterior
TrA	Transversus abdominis

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1 INTRODUCTION

Locomotion is a daily need for humans. To be able to perform movements we need muscles, neuromuscular system and the control of balance (Shumway-Cook & Woollacott 1995, 119. Balance regulates our standing, walking, reaching, running etc. It has been mostly studied within older people because falling is one of the biggest factors for injuries. Fallings and slipping within elderly people can even lead to the death. This makes the control of balance even more important to all humans and an interesting topic among researchers.

Postural control requires a balance control which can be divided into static and dynamic balance. Both static and dynamic balance has been widely studied within researchers. Static balance can be described as a quiet stance where a small sway happens constantly. This is due to constant changes of the center of mass (COM) which is causing different acceleration to different horizontal directions. To be able to keep the balance people regulates their COM inside the base of support (BOS). In a dynamic balance situation, the body's placement is changing and causing the change in foot center of pressure (COP). Many researchers have been studying the relative displacement of COM and COP to reveal different postural strategies. (Enoka 2008, 47, 114; Terry et al. 2011.) Winter (1995) used a model called inverted pendulum model to describe and explain better the control of balance and quiet stance.

The role of arms and upper body has been less investigated than the role of lower body during random balance perturbation. It has been shown that the ankle segment is the first to act to perturbation. The muscles around ankle (TA, SOL and gastrocnemius) have a key role in equilibrium in ankle segment, controlled by central neural system (CNS) and proprioseptors (Winter 1995; Peterka 2002; Woollacott & Tang 1997; Masani et al. 2013; Giulio et al. 2009). In top of the ankle segment, unperturbed balance requires the solid use of hip segment (Matjacic et al. 2001) where the abdominal muscles (TrA) acts first followed by posterior trunk muscles while balance perturbation (Tokuno et al. 2013). The aim of this study is to examine how upper body strategy occurs and especially how arms and deltoid muscle are involved to the control of balance during random perturbation.

2 STATIC AND DYNAMIC BALANCE

For locomotion we need balance. Locomotion can be walking, running or standing and in all of these we need our balance and systems that control posture (Shumway-Cook & Woollacott 1995, 119). To be able to keep the balance, human body has to regulate the body's position, find the point where the mass of the system (body) is evenly distributed in other words center on mass (COM) (Enoka 2008, 47, 114) and keep controlling COM within the base of support (BOS). Physiological three key inputs for the control of balance are believed to be somatosensory, vestibular and visual systems. In top of these systems, balance is also understood and investigated by complex motor control hierarchies. We use our feedback receptors (proprioseptors), muscles, complex multi-segment system, ankle-knee-hip strategies, central nervous system (CNS), reflexes, neuromuscular system and gravity for unperturbed standing. (Winter 1995; Peterka 2002; Woollacott & Tang 1997; Qu & Nussbaum 2009; Matjacic et al 2001; Nashner 1976; Di Giulio et al. 2009; Imagawa et al. 2013.) Adults and children have similar ability to select the right balance strategy (Hatzitaki et al. 2002). In lower body segment, ankle muscles have a major role in maintaining the COM in equilibrium, while upper body segment uses trunk and arm muscles (Masani et al. 2013; Giulio et al. 2009). Even though large amount of knowledge exists, some of the mechanisms for the balance control are still unclear for researchers.

Balance is divided into static and dynamic balance. Static balance describes a situation where humans COM is moving and the surface is not. In dynamic balance both the BOS and COM are moving. This adds the complexity for the strategy we use for maintaining the balance compared to static balance. Dynamic balance requires more than just returning the COM within the BOS. Body's placement changes, foot center of pressure (COP) has bigger role and neuromechanics must accommodate changing support conditions. Relative COM and COP displacements reveal dynamic postural strategies. (Terry et al 2011.)

2.1 Postural control

Postural control has been investigated a lot in a biomechanical field. When the center of body mass is not kept within the BOS a fall occurs (Shumway-Cook & Woollacott 1995, 122). Mostly this is relevant for aging people since falls can sometimes lead to handicap, hip injury and even death, though dynamic balance impairs more than static while aging (Piirainen et al. 2010). To understand and measure postural control different kind of tests, measurements and scales has been studied, compared and created (Berg et al. 1989; Paksuniemi & Saira 2004; Pickerill & Harter 2011; Furman et al. 2013; Qu et al. 2007).

Postural control includes multiple sensory pathways from our body; it includes neuromuscular and motor control mechanism, vestibular system and it occurs both passively and actively (Shumway-Cook & Woollacott 1995, 121; Morasso & Sanguineti 2002).The vestibular system most likely provides the key inputs through the otolith organs by using the orientation of human head with respect to the gravity (Paloski et al. 2006) and position of the head (Johnson & Van Emmerik 2012). Thus, we do know that visual, skin and proprioceptive receptors inside the ear have an important role for regulating the balance. Proprioceptive receptors are located inside inner ears utriculus containing sense organs. These organs are basically hair cells with small stones and fluid-filled tubes in it, which respond to the movement of the fluid and provide the information to the brains which way the head is moving based on the gravity. (Niensted et al. 2009, 486-487; Sawatzky 2009, 58.)

While quiet standing a small sway happens constantly (Fig 1.). The regulation of posture requires a control of position and acceleration of head, arms and trunk (HAT) in the horizontal direction (Winter 1987). The COM is located slightly ahead of the ankle joint and is changing all the time while standing quietly. COM shifting in a relation to the COP can be for example 0,8 cm (Winter 1995; Winter et al. 2003). The sway is highly correlated with ankle joint rotation, which explains why muscles crossing the ankle joint are able to maintain the upright standing position by providing the necessary sensory information

(Loram et al. 2005). There are three general distinct hierarchical levels to help the postural control in a different situations; 1) lower level which includes the muscle tendon units around the ankle joint, 2) intermediate level of control with muscle spindles and Golgi tendon and 3) higher level control which involves the sensory information (van Soest & Rozendaal 2008).

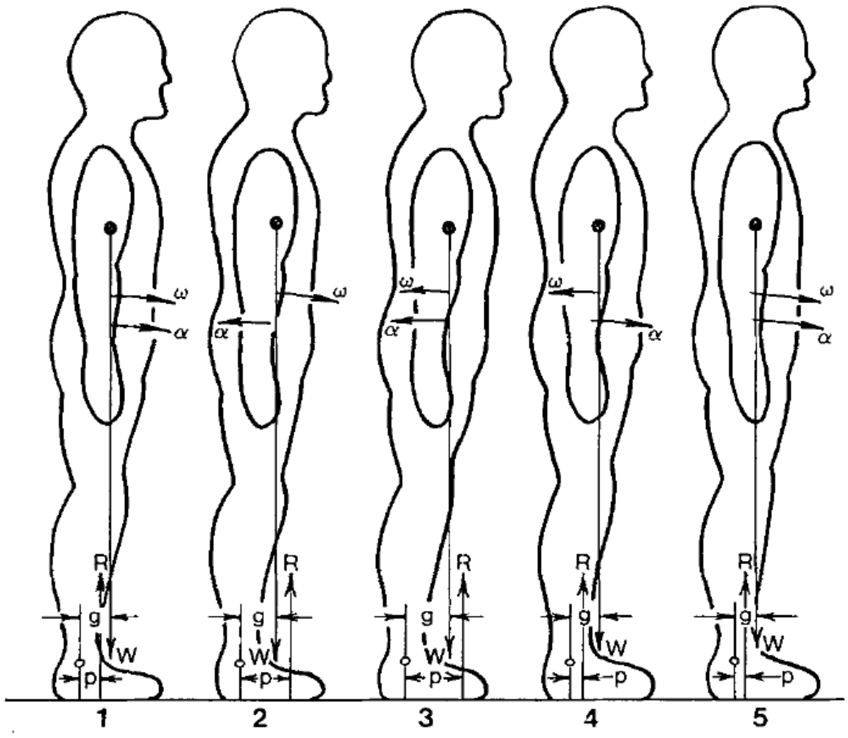

FIGURE 1. A humans body swaying backward and forward while standing (Winter 1995). At point 1 the COM (marked as a dot in top of humerus) stands ahead of p (center of pressure =COP) causing a sway forward. At point 3 acceleration and velocity changes so that sway occurs backwards. g=center of gravity, p=center of pressure, W=body weight, R=ground reaction force, α =angular accelerations and ω =angular velocities.

Golgi tendon organs are sensitive to changes in muscle tension and spindles are sensitive to changes in muscle fiber length (Jones & Round 1992, 68-70). Human body has about

27 500 muscle spindles and highest density is located in neck and hand muscles, because of the eye-head coordination (Enoka 2008, 251). Fusiform shaped spindle fibers are referred to as intrafusal and extrafusal fibers where the first ones, located inside the capsules surrounded by a connective tissue capsule are smaller and are divided into two types depending of the arrangement of the nuclei, contraction speed and the motor innovation. (Enoka 2008, 251-252; Shumway-Cook & Woollacott 1995, 52.)

Imagawa et al. (2013) concluded that the postural control is achieved by synergistic co-activation, and based on the findings the muscle synergies can be investigated using COP fluctuations. Whilst sway, a postural control, COP is moving forward and backward direction. Again, this changes the muscles length (ankle) which is sensed by muscle spindles (Shumway-Cook & Woollacott 1995, 125). Sensitivity of the spindles is modulated by central input via gamma motoneurons which connect exclusively to intrafusal muscle fibers (Enoka 2008, 251). With the changes in the length of the muscles, Ia-afferent neurons are activated with the velocity of 40 – 90 m/s (Enoka 2008, 250) to modulate the movement (Jones & Round 1992, 70-73; Macefield 2005).

Different kind of balance control models has been studied. Maurer and Peterka (2005) applied a model on a proportional, integrative and derivative (PID) by using mathematical models which are challenging to use. Some researchers have been comparing simulated and experimental relationships between sway amplitude and effective stiffness (Winter et al. 1998). Qu and Nussbaum (2009) presented a balance control model which was based on an optimal control strategy to simulate sway behaviors.

2.2 Inverted pendulum model

Inverted pendulum model describes our body while quiet standing, pivoting about the ankle (Fig. 2). By maintaining the COM inside the boundaries (inside COP) is causing slight movement in anterior – posterior (AP) direction. Horizontal acceleration and COP-COM relation while quiet stance, provides valid information about the stability. The main agonist to pretend forward toppling of the body are plantar flexors of the ankle (soleus and gastrocnemius), where antagonist are the dorsi flexors (tibialis anterior). Agonist muscles oppose the toppling torque due to gravity which can be defined as the change of ankle torque per unit change of ankle sway. The ankle stiffness must be at least as large as gravitational driven stiffness in order to remain the system stable. (Winter et al 1998; Bottaro et al. 2005; Gage et al. 2004.) In order to provide stability there has to be joint torque (Runge et al. 1999; Peterka 2002) which is provided by the muscles.

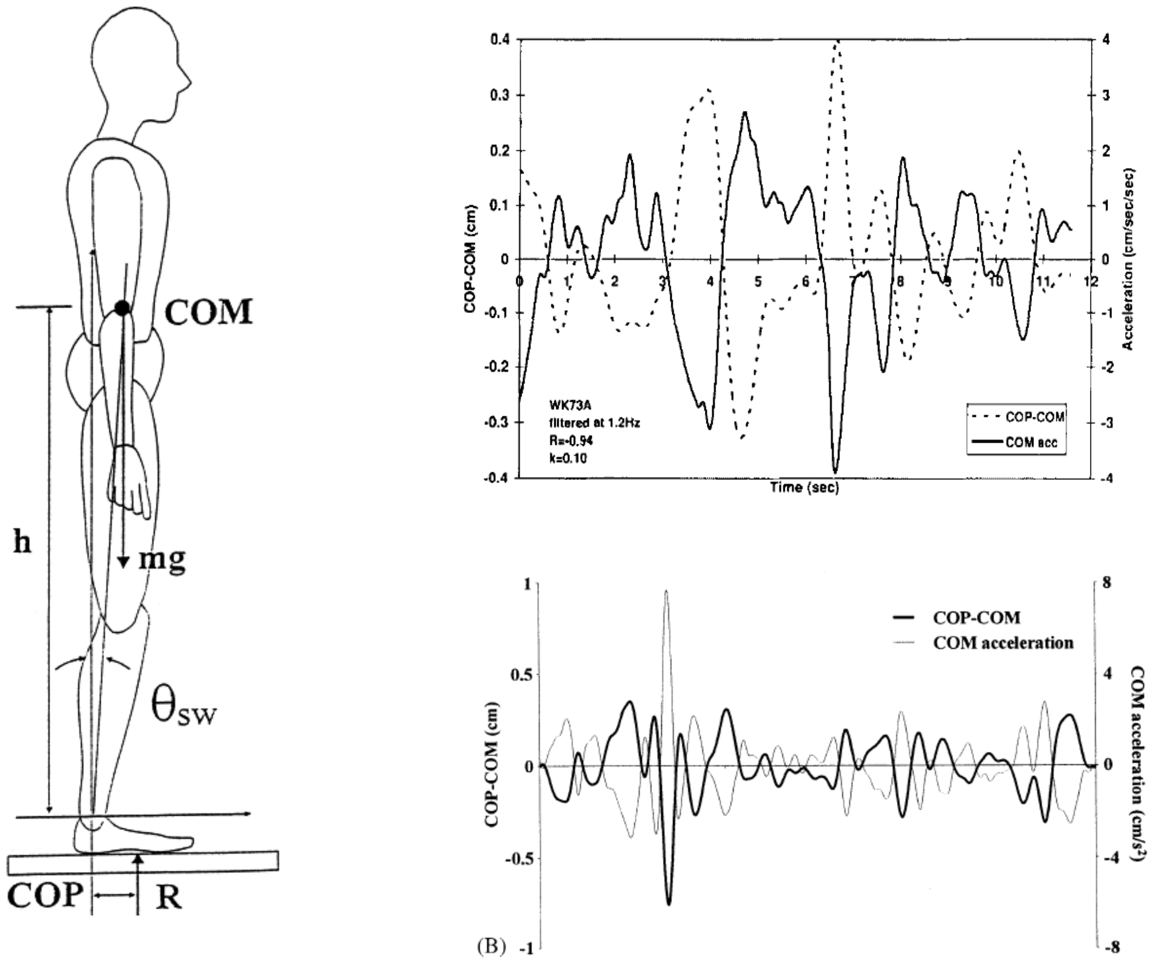

FIGURE 2. Inverted pendulum model in the left (Winter et al. 2003), where body sway (Ɵsw) is seen in a relation of COM and height (h) above ankle. R describes the vertical ground reaction force. Right side shows the high correlation between COP-COM and horizontal acceleration of COM in the AP direction (Winter 1995; Gage et al. 2004) while standing. Upper graph describes COP-COM relations and COM acceleration over 12 seconds.

Besides the ankle strategy also a hip strategy takes a role in inverted pendulum model. Winter (1995) found out in his research that when the platform moved forward, body used tibialis anterior – rectus femoris – abdominalis -strategy. He also stated that CNS recognizes the need to stabilize the joint closest to perturbation first in, and then followed by the knee, hip, and spine. When the ankle muscles cannot participate, an alternative strategy = hip strategy takes place with strongly acting hip flexors (abdominals and rectus femoris).

(Winter 1995.) To achieve balance in more difficult conditions a step strategy has also been mentioned in literature as a third strategy (Kejonen 2002).

3 BALANCE MEASUREMENTS

The measurements of balance and postural stability is beneficial to determining predictors to performance, evaluating musculoskeletal injuries (Herrington et al. 2009), establishing efficiency of physical training and rehabilitation techniques (Sell 2012). The motions of the body and the ground reaction forces with platform translations can be captured in balance control studies (van der Kooij et al. 2005). Balance measurements can be defined in three categories, whereas kinematics measurements are based on analyzing the movement, kinetic measurements are measuring forces and electromyography (EMG) measurements are measuring the muscle activity (Kejonen 2002). According to Müller et al. (1991) to be able to understand better how CNS achieves compensation for body perturbation the EMG analysis with movable platform system is a good measurement technique to use. Many different techniques and tools for upright stance and perturbation can be described with *posturography*.

3.1 Posturography

Posturography describes generally all tools used for measuring human posture or balance which can be controlled by the experimenter. Subject's response to what-ever intervention can be followed and analyzed by using the time-to-peak or peak acceleration, peak velocity, amplitude of the support surface displacements (Visser et al. 2008), root mean square (RMS), sway velocity or standard deviation (Enoka 2008, 201). Typical posturography measurement outcomes are COP, moments and torques from force plate (kinetic), joint angles from Visual 2/3D camera system (kinematic) and muscle activity level from surface EMG as seen in table 1 (Park et al. 2012). Some researchers use intramuscular electrodes but more often surface electrodes are used because it is cheaper and do not require any needles involved (Hermens et al. 1999). Combinations of posturography measurements are preferred within researchers (Visser et al. 2008).

TABLE 1. Typical posturography measurements with outcome measures (Visser et al. 2008).

Modality	Recording Equipment	Outcome measure
Kinetics	- Forceplate	- COP - Torque - Shear forces - Moments
Kinematics	- Optoelectronic 3D camera system	- COG - Joint Angles - Base of Support - MTU length - Segment motion
Electromyography	- Surface Electrodes - Intramuscular Electrodes	- Gross muscle activity - Muscle region activity - Single motor unit activity

3.1.1 Kinematics and kinetics

Kinematics of a body can be measured using motion sensors (markers) and camera systems. By this, the placements of BOS or joint angles are possible to define. Most of the researchers are using different kind of optoelectronic 2D or 3D camera system to measure what is the linear displacement during a trial. Direct measurement of COM can be difficult, and sometimes a single marker can be also placed on the lumbar spine and tracked as an estimated COM position. (Tokuno et al. 2008; Visser et al. 2008.) Recommendations for defining a joint coordination and knowledge of anatomical landmarks are helpful when placing the markers in human body for kinetic measurements. By using recommendations and adapting standards will lead to better communication among researchers. (Wu et al. 2005.)

Kinetic data is providing information about the torques, forces etc. Force plate is used for measuring COP, which is a kinetic measurement of the location of the ground reaction force vector. Normally ground reaction force is the central point of the foot pressure which lies somewhere between the two feet. Newton's law of action-reaction help to define the ground reaction force which is the force provided by the support surface. Ground reaction force is calculated from three dimension / components; vertical (up-down), forward-backward and side-to-side, which the person has transmitted through the feet to the ground and that corresponds to the acceleration. (Enoka 2008, 56-57, 60-61.) Force plates are in many cases integrated to the platform where stance or perturbation is occurring.

3.1.2 Perturbation measurements

Human has optimal strategy control to stimulate sway behavior, perturbation and standing posture (e.g. Qu et al. 2007). A rapid stepping is one of the most natural defense against proprioceptive perturbation (Mansfield & Maki 2009), elevating or lowering response with rapid touch down is great solution while a trip perturbation (Kagawa et al 2011) to avoid falling. Perturbation in balance can be caused by inside from internal (sensory) or outside from external (mechanical) input. While perturbation reflexes, CNS and receptor systems are being used. Static balance is more stable than dynamic while perturbation. This explains why a moving platform (BOS) is closer to the real world –situation (Broglio et al. 2009) and why it has been investigated a lot. Terry et al. (2011) discovered that it is likely that COP has an influence to the chosen strategy differences, because COP reflects the pattern of force application that is not detectable by tracking body movements. In their study they investigated a relative COM and COP displacements and compared dynamic and static balance, and used translations distance of 0.12 m with different velocities (Terry et al. 2011). Corbeil et al. (2013) provided perturbations with surface translation, where a motor-driven platform was moving 0.09 m forward with acceleration of 1.0 m/s2. Translation distances and peak accelerations varies within studies (Weaver et al 2012; McIlroy & Maki 1995; McIlroy & Maki 1999; Tokuno et al. 2013; Piirainen et al. 2013).

Many researchers are trying to find out the muscle activation strategy while perturbation by using EMG. Loram et al. (2005) concluded that COM lags 100 – 300 ms behind muscle activity. Ankle muscles are activating approximately 250 ms after perturbation (Jacano et al. 2004) and arms are showing to initiate 80-150 ms (McIlroy & Maki 1995), where shoulder muscles are turning off approximately 350 ms after perturbation (Weaver et al. 2012). Based on Kagawa and companies (2011) studies, in a case of a sudden slipping, recovery starts in a few hundred milliseconds. Skotte et al. (2004) investigated changes in reactions to sudden unexpected loading and stated that the increase in the average EMG amplitude occurs 50-250 ms after sudden loading. McIlroy & Maki (1995) measured EMG over 100 ms window following the initial onset of perturbation and Winter (1995) observed latencies of 100 – 120 ms in gastrocnemii and hamstring muscles when platform moved backwards. Based on previous studies the voluntary EMG activity takes place somewhere between 80-350 ms. According to many researches older people are unable to initiate movements as rapidly as younger ones, which explains the big window between EMG recordings. (Jacano et al. 2004; Weaver et al. 2012; Mansfield & Maki 2009; King et al. 2009.) However, Tokuno et al. (2010) concluded that greater kinematic differences was found, but not muscle (EMG) activation differences when they compared long and short acceleration-deceleration interval between younger and older adults by surface translation.

Different kind of measurements has been used for perturbation in both standing and seated conditions (e.g. Bjerkefors et al. 2007). Most interesting standing ones lately have been weight-drop cable-pulls (CPs), motor-driven surface-translations (STs) (Piirainen et al. 2013; Weaver et al. 2013; Egerton et al. 2011; Mansfield & Maki 2009; Skotte et al. 2004), loading the subject’s body (Rosker et al.2011; Qu & Nussbaum 2009) or pushing it forward (Kim et al. 2012) and tilting and / or rotating the surface (Goodworth & Peterka 2009). St-Onge and colleagues (2009) studied upper body and suggested that when the translation of the platform of the chair occurs forward, neck and trunk muscles are activating first, whereas for backward translation, extensor muscles are activating first followed by flexors with healthy subjects.

The direction of the platform translation has an impact to the use of balance control (Preuss & Fung 2008) but the predictability of the translation direction don’t have an influence to the upper body strategy or abdominal muscle recruitment order (Tokuno et al. 2013). Preuss and Fung (2008) reported that when translation of the surface direction occurred forward the upper body displacement was approximately 20 mm (HAT COM), comparatively when translation was backward displacement was less than 10 mm. Piirainen and colleagues

(2013) observed similar results and concluded that forward translation showed more evident in balance control than backward.

Some studies have been showing that the adaptation to sudden and unexpected loading to the trunk occurs after first couple trials (Skotte et al. 2004). Schmid and colleagues (2011) found similarities with learning patterns when they demonstrated the aim of the CNS to keep COM within limits in different conditions; eyes open, eyes closed, high and low frequencies. They reported that in slight perturbations there was not a major activation of gastrocnemius and soleus muscles which are in line with the human’s optimal trade-off between task-level performance and minimizing energy expenditure (Schmid et al. 2011).

4 FACTORS AFFECTING BALANCE

There are many factors affecting to balance control. Recovery from sudden perturbation is a multi-joint task for body, where learning and exercise history has also an impact not forgetting the age factor and neurological, vestibular and pathological issues such as Parkinson's disease, strokes, multiple sclerosis (Terry et al. 2011). Skotte et al. (2004) investigated changes in reactions to sudden unexpected loading and found out that muscle reaction was much slower in first two trials (468 ms) compared to trials 3-10 (365 ms) which indicates that learning is affecting to the balance control and measurements. Paksuniemi and Saira (2004) found out in their studies that athletic humans have better and faster strategies for recovering from perturbed situation than non-athlete humans, especially judokas when comparing to other sports athletes. Influence of alcoholism has also been studied. Alcoholic men and women have longer sway paths and difficulties stabilizing quiet stance compared to healthy population (Sullivan et al. 2010). Fatigue can influence to humans balance control, though there is no direct evidence for it (Fuller et al. 2011). Madigan et al. (2006) found out that when subjects were suffering from fatigue, they adopted a slight forward lean position and also they observed changes in sway which can be seen as increased joint angle variability at multiple joints.

4.1 Control strategies

4.1.1 Role of ankle and hip segments

Small perturbations are often handled by ankle strategy (Nashner 1976; Winter 1995), which is defined as a single-segment inverted pendulum about the ankle joint, and big perturbations are handled by multi-segment strategies. These strategies generate a precise mechanism of corrective torque, same as a moment of a force which describes the capability of a force to produce a rotation (Enoka 2008, 45). Multi-segment strategy is a closed-loop process where CNS affects both the output (joint torques) and input signals (joint angles).

Perturbation is needed to “open” the loop. (Shumway-Cook & Woollacott 1995, 457; van der Kooij et al. 2005.)

Several studies have been done to define which parts of body influence and how to the balance control while perturbation. Use of upper body, hip flexors, as a segment creates a rotational motion that occurs about the L4/L5 spinal joint area. Matjacic et al. (2001) have been investigating ankle and hip correlations while Goodworth and Peterka (2009) studied the orientation of the upper body relative to the pelvis while lower limbs and pelvis were held in a fixed position. In all studies it has been clear that the CNS has main role for the recovery of perturbation. In most of the studies the strategy of proprioceptive recovery has been investigated mainly in 2 orthogonal directions; anterior - posterior (AP) and / or medialis – lateralis (ML). (Matjacic et al. 2001; Winter et al 2003; Winter 1995; Goodworth & Peterka 2009; Imagawa et al. 2013; Sullivan et al. 2009.)

Runge and colleagues (1999) concluded that joint torques, which indicate how body movements are produced, are useful in defining postural control strategies. They examined the hip strategy in postural responses to backward translations, and found out that based on the translation velocity (5 cm vs. 55 cm), a mixed hip and ankle strategy was introduced, where faster translations revealed the addition of a hip flexor torque to the ankle plantarflexor torque (Runge et al 1999).

Soleus (SOL) muscle is the main lower body agonist regulating quiet standing with conditions based on Di Giulio and colleagues (2009) findings. To be able to exert a force, each degree of freedom about a joint is controlled by another muscle which is providing opposing actions (Enoka 2008, 298). In this case the antagonist muscle would be tibialis anterior (TA) muscle. It top of SOL and TA, a gastrocnemius is taking a big part of ankle strategy acting closely like soleus. Soleus and gastrocnemius are attached to the heel with Achilles tendon. Recent evidence shows that TA may in fact exhibit favorable behavior for proprioception during balance control (Di Giulio et al. 2009).

4.1.2 Role of upper body

Upper body uses different muscles for the balance control. Muscles in the anterior and posterior side of trunk together with arms, neck and head creates the upper body segment. Preuss & Fung (2008) concluded that displacement of trunk COM was significantly smaller in standing than in sitting when they investigated upper and lower body perturbation strategy and directions differences.

Rectus abdominis (RA) is one of the abdominis muscle group which is believed to be the main muscle group to determine postural control. Abdominal muscle recruitment strategy has been studied by many researches. Recently Tokuno et al. (2013) found out that the predictability of the translation movement (forward or backward) does not effect to the abdominal muscle recruitment order, neither does the unexpected loading (Cresswell et al. 1994). Transversus abdominis (TrA) has been considered to be the primary responsible muscle for maintaining stability. Other abdominis muscles involved to equilibrium are the superficial ones (obliquus). (Tokuno et al. 2013.)

Erector spinae (ES) muscle is located in the back of the trunk and has important role for trunk extension and maintaining humans' postural control (Nienstedt et al. 2008, 149-150). As well as abdominal muscle group the ES controls the equilibrium and is one of the three main muscles which are considered to be the primary balance-correcting muscles during a backward surface translation (Tokuno et al. 2010).

Deltoideus muscle is in the shoulder, primarily moving the arm and good point for measuring arm reactions (King et al. 2011). Since its shape as a triangle, many researches use both anterior (AD) and posterior (PD) side of it in EMG measures. Measurements where the surface translations direction is forward, causes body to swing backward. In this case, the largest EMG responses in result will be measured from the PD. Conversely, AD is acting mostly while surface is moving backward and body is swaying forward. (McIlroy & Maki 1995; Mansfield & Maki 2009.)

4.2 Neuromuscular properties

4.2.1 Ageing

Age has probably the biggest influence to the balance, particularly the loss of muscle forces in lower limbs (Pajala et al. 2008, 137). Ageing makes all movements from reflexes to multitask activities slower (Enoka 2008, 401). Based on researches maximal voluntary force is greatest at the age of 20-30, during and after fifth decade muscle mass starts to decrease as well as explosive force due to neuromuscular system. The loss is more rapid with women than men; cause is probably the hormonal balance after the menopause. (Jones & Round 1992, 95-96; Häkkinen et al. 1995; Häkkinen 1994; Frontera et al. 1991.) Koceja et al. (1999) studied static conditions between young and elderly and found out that the young subjects produced significantly less postural sway (young; 3.80 mm vs. elderly; 4.89 mm). Piirainen and colleagues (2010) found out that re-stabilization and rapid force production after sudden disturbance is age related when they measured maximal isometric torque and activation levels from lower body muscles in a group of younger ones (age 21-31 years) compared to the older ones (age 60-70 years). On the other hand, there is evidence showing that the foot-off time is smaller with older population even though the length of the step taken while surface translation is similar with older and young adults as seen in table 2 (Mansfield & Maki 2009).

TABLE 2. Examples of previous surface translation studies showing age-related differences in characteristics of stepping reactions (modified, original Mansfield & Maki 2009). OA stand for older adults and YA stand for younger adults.

Study	Unpredictability			Instruction	AP-step measures			ML-step measures	
	Onset timing	Magnitude	Direction[a]		Foot-off time	Swing duration	Step length	Cross-over steps	Foot collisions
Surface-translation (ST) perturbations:									
McIlroy and Maki (1996)	Yes	No	Yes (F, B)	Try not to fall	OA = YA	OA = YA	OA = YA	–	–
Maki et al. (2000)	Yes	Yes (‘low’ and ‘high’ magnitude)[d]	Yes (L, R, F, B)	React naturally	–	–	–	OA = YA[e]	OA > YA[e]
Present findings	**Yes**	**Yes (CPs included)[d]**	**Yes (L, R, F, B)**	**React naturally but minimise number of steps**	OA < **YA**	**OA = YA**	**OA = YA**	**OA = YA[e]**	**OA > YA[e]**

The length of the platform translation has been suggested to have an impact to the use of balance control with elderly people. Tokuno and colleagues (2010) reported that bigger kinematic differences in postural responses was found in older people (age 66-81 years) when the translation of the surface was long compared to short translations. Younger adults were aged between 22-39 years old in this study, where researches did not find any change in EMG latencies or amplitudes between two groups (Tokuno et al. 2010).

4.2.2 Visibility and head conditions

Visual ability influences to the balance control, or does it? Fransson et al. (1998) suggested that eyes-open condition provided smaller amplitude in sway compared to eyes-closed situation (young eyes open (EO) 3.80mm vs. eyes closed (EC) 5.44mm; elderly EO 4.89 mm vs. EC 5.95 mm). The visual availability has only a small effect on the upper body sway (Goodworth & Peterka 2009). On the other hand, Di Giulio et al. (2009) collapsed and averaged eyes-open and eyes-closed data together because they did not observe any differences between these conditions in lower body (TA, SOL and gastrocnemius) study. Head condition with visibility has been also studied. Agostini and company (2013) observed volleyball players' postural sway with 10 different head conditions and found out that the defensive players, whose role requires the quickest reaction time, differed from control while eyes open but not significantly with eyes closed. Paloski et al. (2006) discovered that subjects were able to maintain upright stance with static tilts with eyes closed, although the degree of postural de-stabilization varied directly with the frequency of the head tilt / neck extensions.

5 THE PURPOSE OF THE STUDY

Based on previous studies, it has been shown that the COM and COP have the key input for measuring balance perturbation. Numerous studies have established that ankle strategy takes place in perturbed balance and correlation with hip strategy has been studied. Based on the literature the abdominal muscles especially TrA reacts first in the trunk when perturbation occurs. But only a few studies have been investigating the arms influence to the balance perturbation. Thus, the main purpose of this study was to extend the knowledge on the hip strategy to the arms; what is the reaction of the arms during perturbation? And moreover, are there strategically changes when using or not using your arms during perturbation?

In this study the COP was used as a trigger and by that the perturbation was randomized. One other purpose of the study was to see the latency of the deltoid muscle activation and compare RA versus ES muscle activation order. It was also interesting to see how the other measured upper body muscles behaves and were the RA and ES co-operating or behaving similarly while perturbation. The main hypothesis of the study was that the hip strategy includes the freedom of arms. When the freedom is taken away and the arms are crossed (No Arms –condition) the strategy changes to involve more trunk muscle activity.

6 METHODS

6.1 Subjects

Five healthy young and middle-aged subjects participated to the study. Four male and one female subjects participated (N=5) as seen in Table 3. All participants were informed about the protocol and safety before the actual measurements. They were also told about their right to retire from the study at any time.

TABLE 3. Subject characteristics with a mean ± one standard deviation.

N	Age	Height	Weight
5	32 ± 11 years (range of 23–50 y)	177 ± 11 cm	77.3 ± 15.7 kg

6.2 Protocol

The study took place in February - April 2013 in Brisbane Australia in the UQ human movement studies laboratory. The subjects were advised to wear sport clothing during the study and no warm ups or no special preparation was needed prior the study. Subject's gender, age, height, weight and handedness were written down. Study was performed standing still, eyes open where the surface under the subject was moving either anterior or posterior (AP) directions providing the unbalanced situation to the subjects. Descriptive statistics were observed and calculated during each condition.

Practice was included to the protocol. There were three times three practice trials to eliminate the learning factor influence to the results and to help the subject to get used to the perturbations. One set of three trials was performed before and one set after the placement of EMG electrodes and markers. Prior to testing, the MVCs were performed by a help of a test personnel as maximum voluntary efforts for EMG normalization purposes. One more practice trial was performed on the force plate / treadmill after the MVCs. A “safety person” was standing behind the subject during the whole study for safety because no harness was used. For repeatability and standardization purposes, each subjects’ feet were positioned in the middle of the force plate where a piece of a tape was visually showing the middle point. During the testing phase, the subjects were asked to stand normally and look at the dot in the wall in the other side of the room. Subjects were also instructed not to take a correcting step if possible. Furthermore, if a corrective step was needed, subject was asked to try to make it as late as possible.

Study protocol included four blocks of 10 trials which were performed with the use of treadmill. There were totally 40 trials. After each 10 trial there was 60 seconds of rest, but if the subject felt fresh, the study continued without the rest.

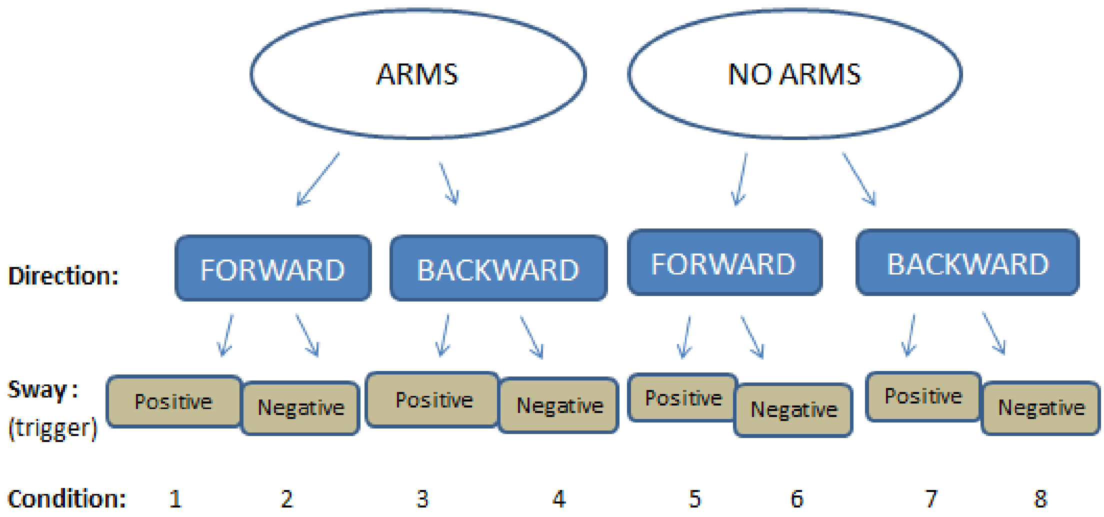

FIGURE 3. Protocol of the study with eight different conditions; arms free or no arms, surface translation direction forward or backward with positive of negative sway as a trigger (COP).

Perturbations were randomized with 8 different conditions with different movement patterns, which all occurred five times (Fig. 3 & table 3). COP was used as a trigger. When a subject was swaying forward it was recorded as a positive sway. In condition ‘No Arms’ the subject crossed his / hers arms in front of him / her abdomen by cross holding his / hers elbows. Treadmill velocities and accelerations changed between the surface translations directions but the distance was always the same 0.2 m (Table 4).

TABLE 4. Treadmill velocities, accelerations, decelerations and distances in forward and backward translations.

	FORWARD	**BACKWARD**
Velocity (m/s)	0.24	1.0
Acceleration (m/s^2)	1.25	1.0
Deceleration (m/s^2)	1.25	1.0
Distance (m)	-0.20	0.20

In the results only measurements from two different conditions are presented. Condition two with arms and condition six without arms, which were both forward direction with negative sway as trigger, were picked for further analysis because there was the highest variation in COP data when comparing all eight conditions. In the condition two the subject was having a free use of his / her arms with surface translation forward direction where the subject’s negative sway was acting as a trigger as seen in figure 3. In Condition six the only difference to condition two was that no arms was used. When the sway was negative it provided more real life situation when slipping. By choosing these two conditions it supported the purpose of this study.

6.3 Measurement equipment and analyses

6.3.1 Electromyography

Preparation included surface EMG electrodes (1.5 cm diameter, Ag/AgCl, Covidien, Mansfield, MA) which were placed according to the recommendations by SENIAM (Hermens et al. 1999) in seven different muscle in the right side of the body; Soleus (SOL), Medialis Gastrocnemius (MG), Tibialis Anterior (TA), Rectus Abdominis (RA), Erector Spinae (ES), Deltoideus Anterior (AD) and Deltoideus Posterior (PD). The maximal voluntary contraction (MVC) was measured before the trials started without any warm up. The subject executed the MVC for TA, SOL and MG by maximal dorsi- and plantarflexion when another person (test crew) was resisting the movement. The MVC for RA was done with situp where another person was resisting the movement by holding hard from the chest. MVC for ES was performed similarly, where the subject was lying in the floor facing down and executing a back extension while another person was resisting the movement from the upper part of the back. AD and PD MVCs were performed while standing and executing an arm flexion and extension with resistance by another person.

Signals were amplified 350-2000 times (MA300, Motion Lab Systems, LA, USA), depending on the signals strength and band pass filtered: low band filter 18 Hz and high band filter 500 Hz with transition gap of 12. Sampling rate was 2400 Hz using Spike 2 data collection system (Cambridge Electronics Design, Cambridge, UK). The raw data was rectified in Spike-system. The averaged EMG amplitudes were calculated as RMS amplitude (RMS amp.) of the EMG signals with 300ms window to minimize the noise. The EMG amplitude data was also taken into further analyze by diving it into three time windows after onset (0 to 50 ms = EMG_{50}, 50 to 150 ms = EMG_{150} and 150 to 300 ms = EMG_{300}, respectively) (Figure 4). RMS amplitude was also used for calculating the maximal EMG (MVC) with 200ms window pointing to the highest value of each signal. The results are shown as normalized EMG (RMS $\%EMG_{max}$) which was calculated by dividing the RMS amp. –value with the MVC-value.

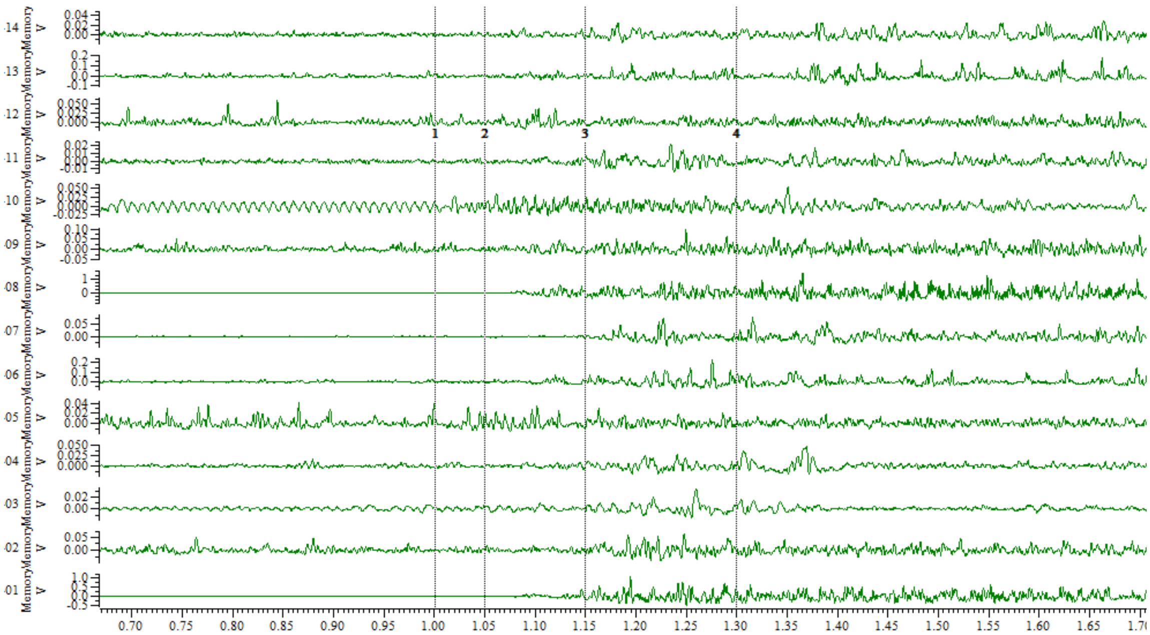

FIGURE 4. EMG measurements with time windows Subject 4. From the top AD, ES, MG, PD, RA, SOL and TA, where channels 1 – 7 presents Arms-condition and channels 8 – 14 No Arms –condition.

6.3.2 Kinetics

An instrumented force sensing treadmill (tandem) with power plate under the belt was used for the measurements (AMTI, Watertown, MA). COP was used to trigger the movement of the treadmill for a perturbation (Winter et al. 1996, 2003). COP was controlled by using custom written software with sampling rate of 2400 Hz (LabView National Instruments, Austin, TX). The force measurements from power plate was taken into further analyze as well as the displacement of COP. Both variables were analyzed in AP (x) and ML (y) directions.

6.3.3 Kinematics

For kinematics measures an eight camera, three-dimension (3D) optoelectronic motion capture was used (Oqus, Qualisys AB, Gothenburg, Sweden). 39 spherical reflective markers (38mm in diameter) were used. They were attached to anatomical landmarks using double sided tape (Dumas et al. 2007; Wu et al. 2002, 2005) as seen in figure 5. The position data was sampled at 200Hz using a specific motion analysis software and hardware (QTM, Qualisys AB, Gothenburg, Sweden) and exported for offline analysis in 3D-software (Visual 3D, C-Motion, Kingston, Canada). Prior to testing, a static standing trial was completed to create the model for limbs. After static trial, 14 of the markers were removed, leaving only the markers and clusters (lower limbs) on the respective segments.

FIGURE 5. Full marker setup with EMG electrodes from anterior and posterior side of the body.

The kinematics were analyzed by looking at the results of the three upper body joints from the right and from the left side of the body; hip, shoulder and elbow. Within all these three joints a three different variables were analyzed; joint velocities, angles and accelerations. Also lower body joints were measured (ankle and knee) with the same variables but those were not analyzed because the interest of this study was the upper body.

6.4 Statistics

The statistical analyses were performed by using IBM SPSS Statistics Version 20. The results are presented by two conditions “Arms” and “No Arms” except the EMG results which are also presented by three different time windows. The RA muscle results are also presented by three different time windows per subject due to findings along the analyses. The between conditions differences were analyzed using the 2-related samples test. Wilcoxon sign-rank test were applied. The significance level was set at $p \leq 0.05$. Relationships between Arms and No Arms –conditions with variables and percentage changes in the joint angles, forces, COP and EMG measurements were analyzed using Spearman correlation coefficients.

7 RESULTS

Electromyography. The averaged EMG (RMS) activity was measured during forward perturbations in totally seven different muscles in free use of arms and No Arms -condition. The EMG activity from anterior deltoid (AD) muscle was 65.7 ± 2.1 % higher ($p < 0.05$) with free use of arms compared to the condition with no arms as seen in figure 6.

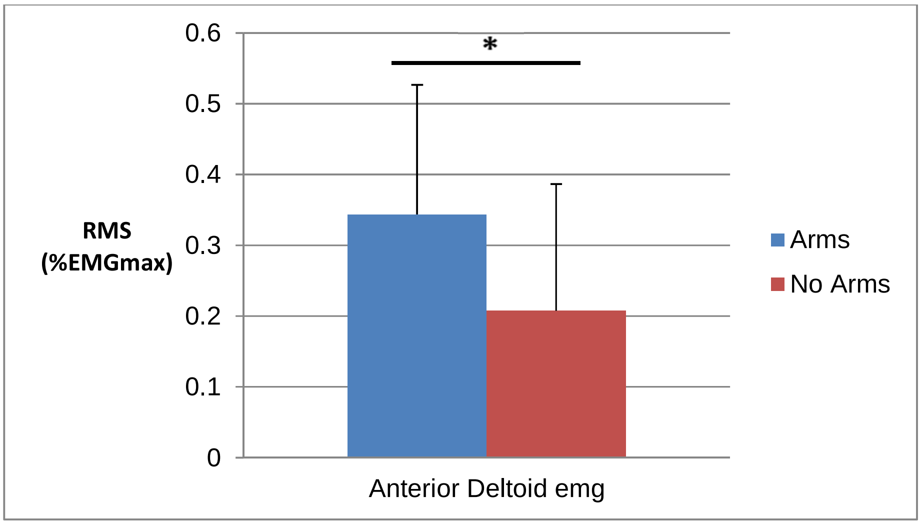

FIGURE 6. EMG results of Anterior Deltoid with two different conditions (Arms and No Arms).

In figure 7 is shown the results of rectus abdominis (RA) muscle. The EMG activity in No Arms –condition was 54.9 ± 13.8 % higher ($p < 0.05$) compared to the Arms-condition.

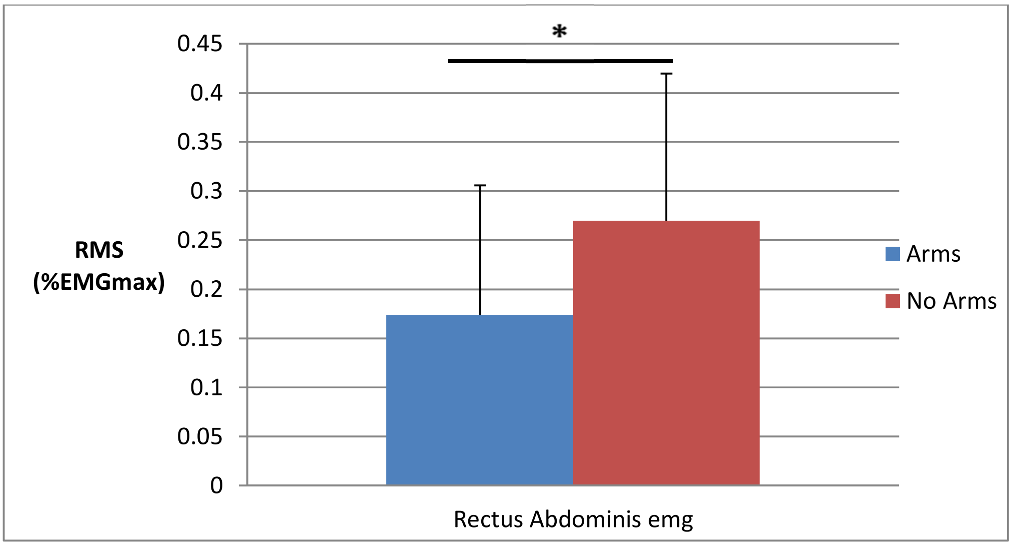

FIGURE 7. EMG results of Rectus Abdominis with two different conditions (Arms and No Arms).

Based on the results no significant (n.s.) difference was found in erector spinae (ES) muscle, but there was a 190 ± 650 % higher EMG activity of the muscle in No Arms –condition compared to the other condition with free use of arms as seen in figure 8.

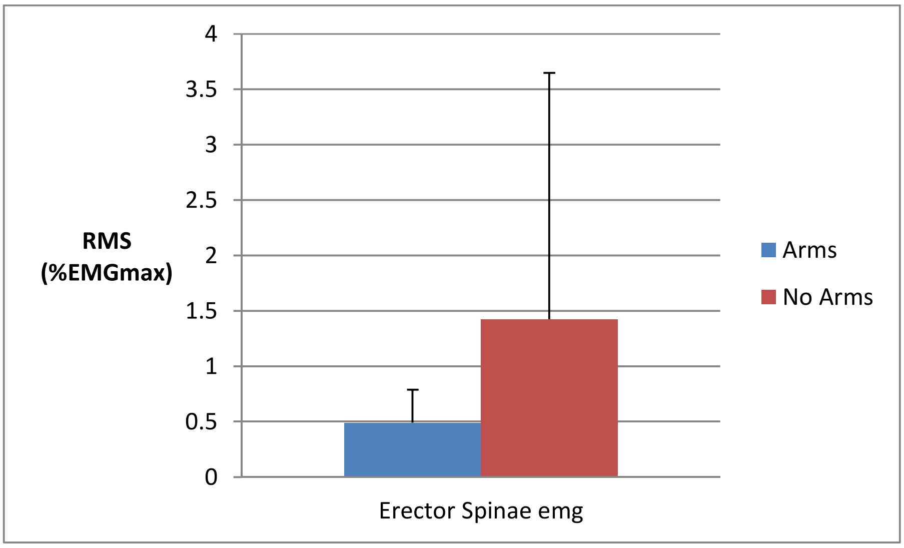

FIGURE 8. EMG results of Erector Spinae with two different conditions (Arms and No Arms).

The EMG activity of posterior deltoid (PD) muscle was 18.2 ± 60 % higher (n.s.) in No Arms –condition (Figure 9).

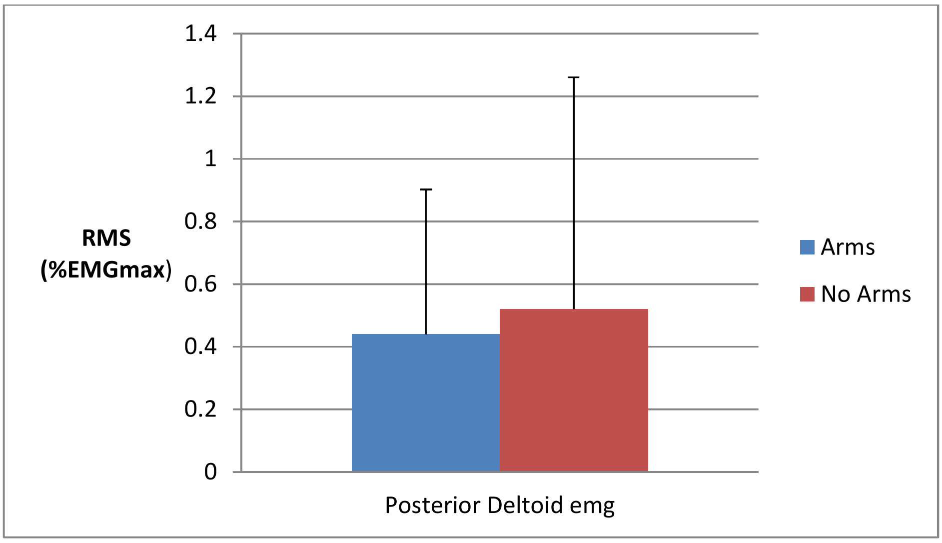

FIGURE 9. EMG results of Posterior Deltoid with two different conditions (Arms and No Arms).

Lower body muscles were also measured in this study. Soleus (SOL) was showing 24.4 ± 20.7 % higher (n.s.) EMG activity and medial gastrocnemius (MG) 27.4 ± 43.9 % higher (n.s.) EMG activity when arms were used as seen in figure 10. The tibialis anterior (TA) muscle EMG activity was 6.6 ± 4.8 % higher (n.s.) when No Arms –condition was used compared to Arms-condition (Figure 11).

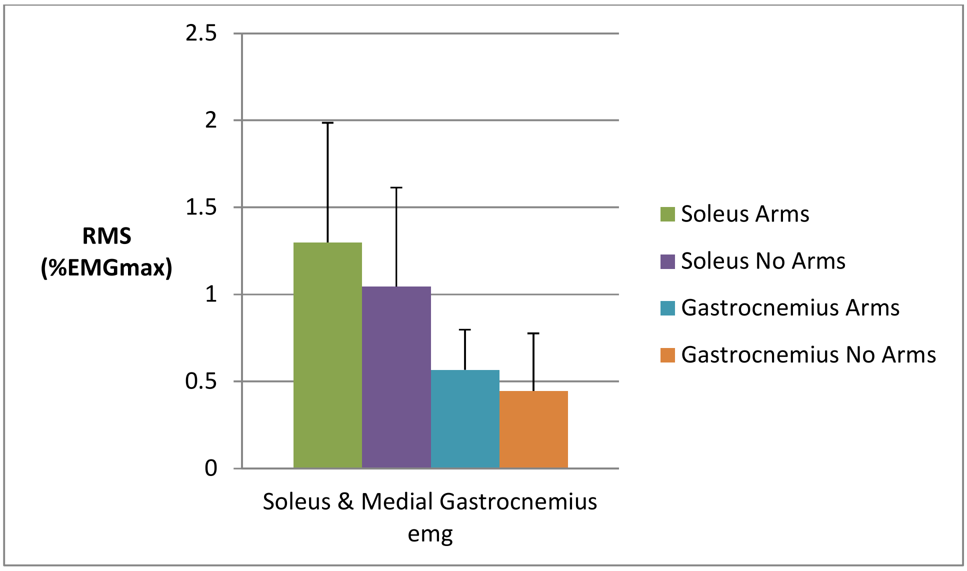

FIGURE 10. EMG results of Soleus and Medial Gastrocnemius with two different conditions (Arms and No Arms).

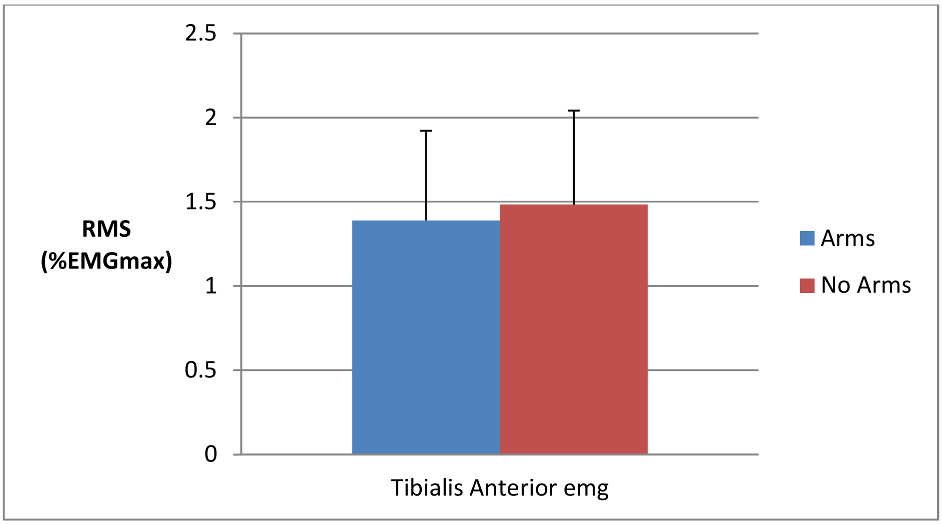

FIGURE 11. EMG results of Tibialis Anterior with two different conditions (Arms and No Arms).

EMG with different time windows from the onset of perturbation. More detailed analyses for EMG was done by dividing the EMG into three time windows: EMG_{50}, EMG_{150} and

EMG_{300}. In RA muscle the EMG activity was higher when no arms were used in all timeslots. Based on the results the RA EMG was 61.2 ± 40.8 % higher (n.s.) in EMG_{50}. In EMG_{150} the RA EMG was 59.1 ± 2.4 % higher ($p < 0.05$) and in EMG_{300} (150 – 300 ms after the perturbation) it was 51.7 ± 18.4 % higher ($p < 0.05$) when No Arms was used compared to the Arms-condition as seen in figure 12. The EMG activity increased by 16.3 ± 10.2 % ($p < 0.05$) from EMG_{150} to EMG_{300} in Arms-condition.

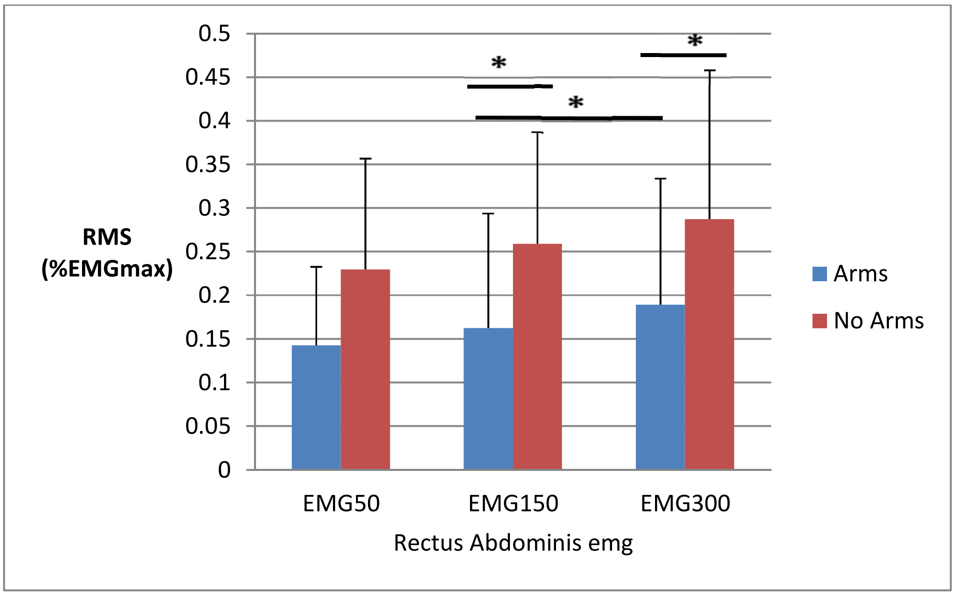

FIGURE 12. The averaged EMG activity in RA muscle with three time windows: EMG50 = 0 – 50 ms, EMG150 = 50 – 150 ms and EMG300 = 150 – 300 ms after the perturbation with two different conditions (Arms and No Arms).

Within all subjects the RA activity was higher when no arms were used. However, there were differences between the subjects and between the timeslots as seen in figures 13 and 14. Subject number 4 differed from the group by having much greater amplitude than the others.

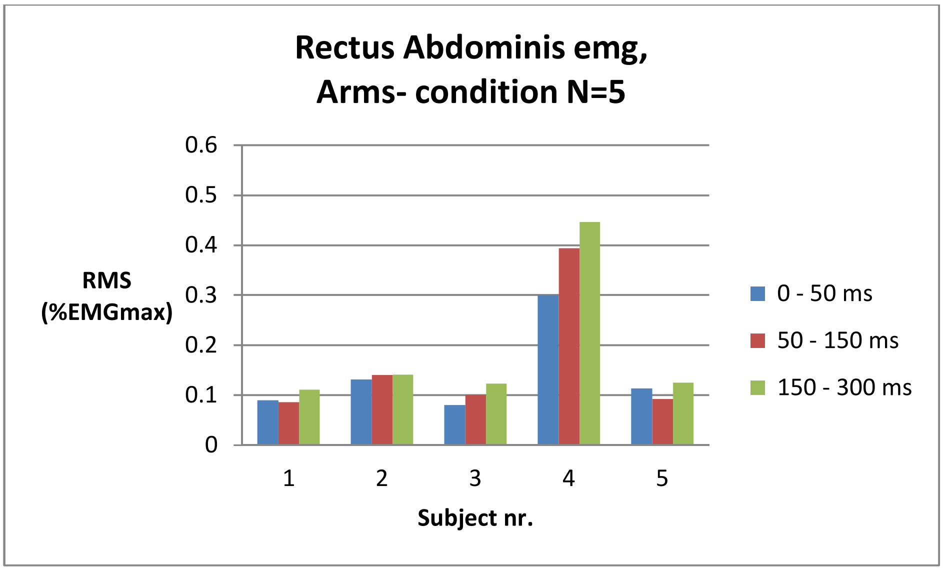

FIGURE 13. The EMG activity in RA muscle in each subject with three timeslots in Arms-condition.

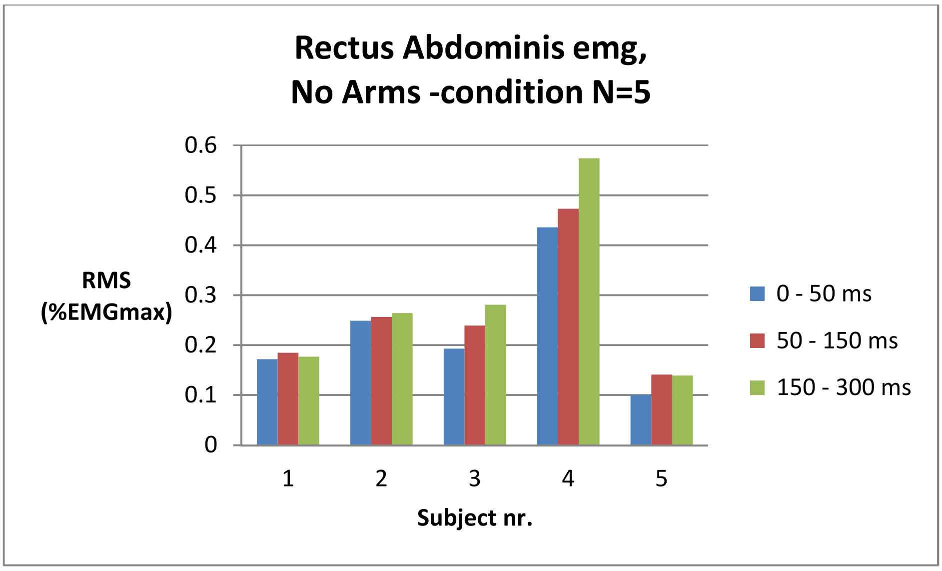

FIGURE 14. The EMG activity in RA muscle in each subject with three timeslots in No Arms – condition.

The EMG activity in AD muscle was more active when having free use of arms in all three timeslots. The AD EMG was 43.0 ± 36.8 % higher (n.s.) in EMG_{50}. In EMG_{150}, the AD EMG was 41.0 ± 16.2 % higher ($p < 0.05$) when using arms and in EMG_{300}, the AD EMG was 83.5 ± 16.1 % higher (n.s.) with arms (Figure 15). Activity level increased from EMG_{50} to EMG_{150} time ($p < 0.05$) in both conditions. Based on the results, when arms were not used (No Arms –condition), the EMG increased by 21.8 ± 19.2 % ($p < 0.05$) from EMG_{50} to EMG_{300}.

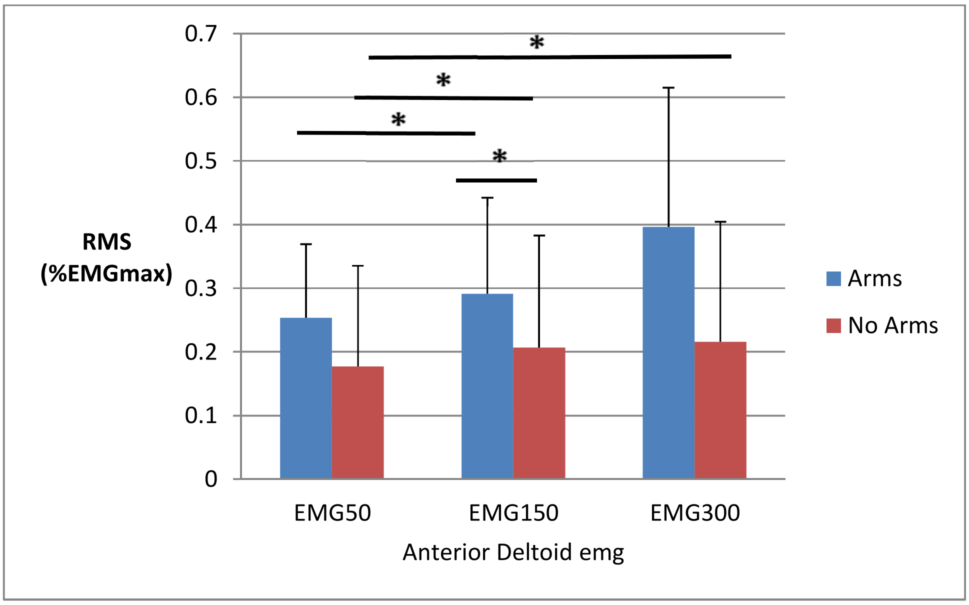

FIGURE 15. The EMG activity in AD muscle with three time windows: EMG50 = 0 – 50 ms, EMG150 = 50 – 150 ms and EMG300 = 150 – 300 ms after the perturbation with two different conditions (Arms and No Arms).

The EMG activity measured from the SOL muscle was 24.3 ± 17.4 % higher ($p < 0.05$) in EMG_{300}, 150 – 300 ms after the perturbation, when using arms (Figure 16).

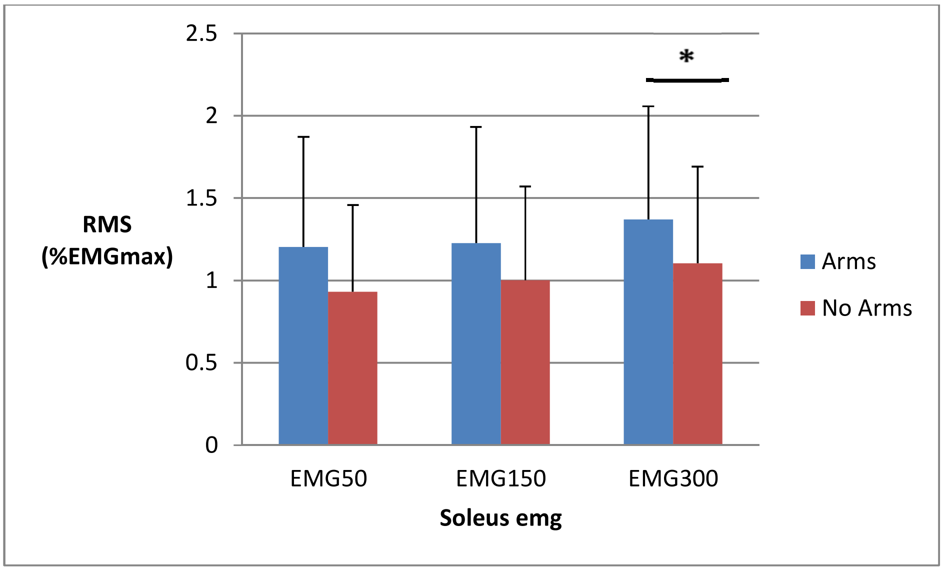

FIGURE 16. The EMG activity in SOL muscle with three time windows: EMG50 = 0 – 50 ms, EMG150 = 50 – 150 ms and EMG300 = 150 – 300 ms after the perturbation with two different conditions (Arms and No Arms).

Forces. Based on the results of the force peak-to-peak-amplitude, the Force_x (AP) was 1.4 ± 34.7 % higher (n.s.) and Force_y (ML) was 12.8 ± 54.0 % higher (n.s.) in No Arms -condition (Figure 17). The COP in the AP direction was 36.1 ± 30.6 % higher (n.s.) and COP ML direction was 38.0 ± 2.4 % higher (n.s.) with free arms compared to the No Arms -condition as seen in figure 18.

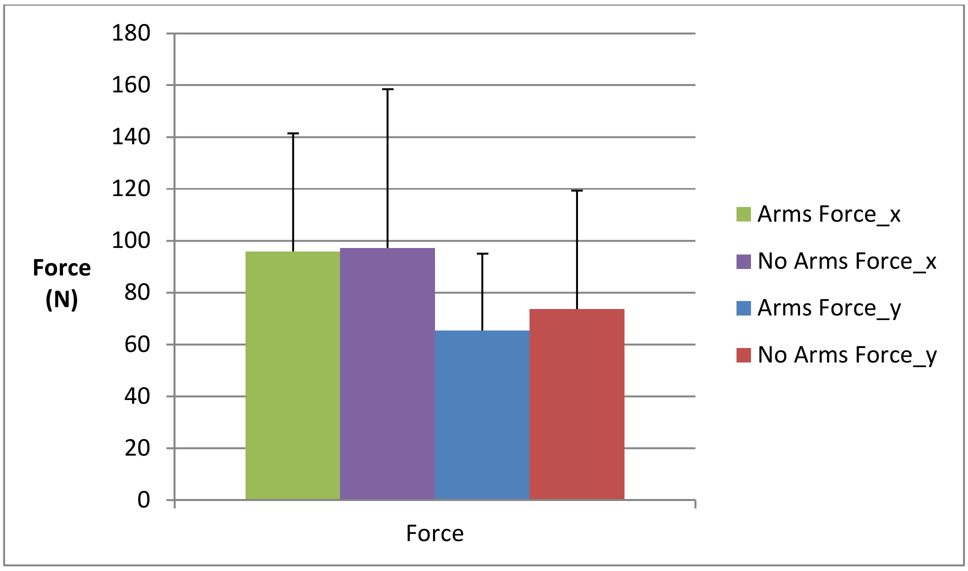

FIGURE 17. Force results x (AP) and y (ML) directions with two different conditions (Arms and No Arms).

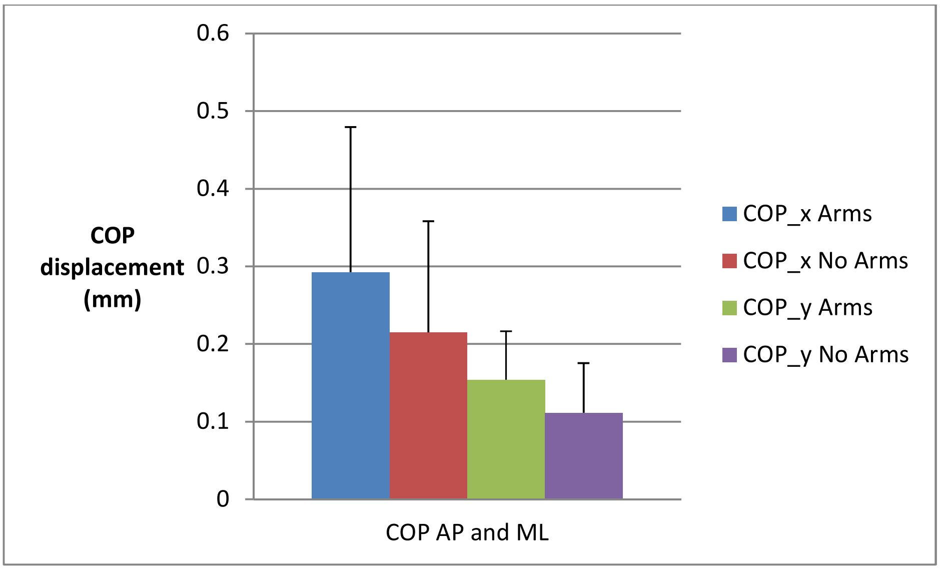

FIGURE 18. COP results x (AP) and y (ML) directions with two different conditions (Arms and No Arms).

Kinematics. The measurements of the upper body joints (hip, shoulder and elbow) velocities, angles and accelerations did not differ significantly between the conditions, whereas the averaged hip acceleration in the AP direction from the right side of the body was 89.6 ± 64.0 % higher (n.s.) in Arms-condition compared to No Arms –condition (Figure 19).

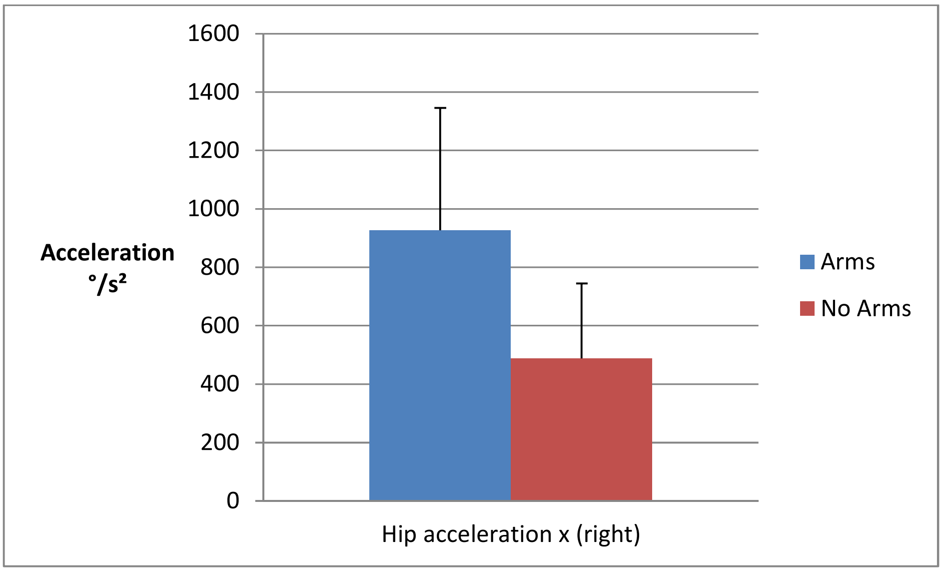

FIGURE 19. Hip acceleration x (forward – backward) direction from the right side of the body with two different conditions (Arms and No Arms).

The angle of the hip was 5.8 ± 24.0 % higher (n.s.) in AP direction in the Arms-condition compared to No Arms -condition as seen in figure 20.

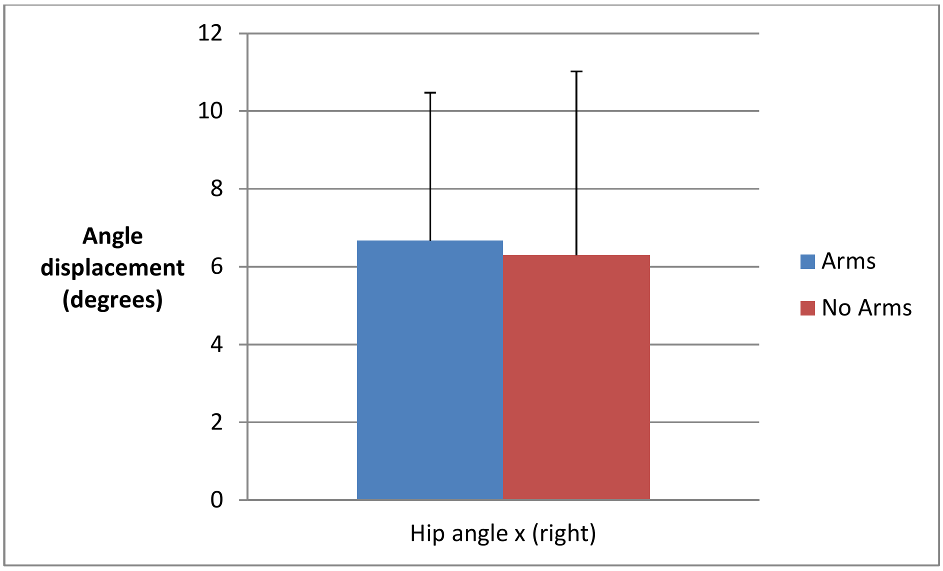

FIGURE 20. Hip angle x (AP) direction from the right side of the body with two different conditions (Arms and No Arms).

8 DISCUSSION

The purpose of the study was to investigate how the use of upper body influences to the human balance during perturbation. The subjects of the study were standing in top of the treadmill and waiting for the perturbation to occur. Perturbation was sudden based on the sway direction of the subject. Platform moved forward and backward direction in the perturbation. There were totally eight conditions which each was performed five times but with randomized order. Based on the highest variation of the COP, two different conditions to forward direction were investigated and taken into further analyzes – Arms and No Arms – where the last one was performed with arms crossed in front of subject's stomach. The main finding is that there is no significant differences in balance control weather you use or not use your arms when perturbation occurs in the forward direction. Nevertheless, to support the hypothesis of this study, a higher EMG activity of the upper body was observed when not using arms (No Arms –condition). ES muscle was showing higher EMG activity in No Arms –condition as well as the EMG of RA and AD muscles had a significant role in the balance control -system when not using arms ($p < 0.05$).

It has been previously investigated that the muscles around ankle (TA, SOL and gastrocnemius) have a key role in equilibrium in ankle segment (Winter 1995; Peterka 2002; Woollacott & Tang 1997; Masani et al. 2013; Giulio et al. 2009). In this study, where perturbation occurred into forward direction and subject had a feeling of slipping backward, the EMG of SOL muscle was 24.4 % higher and the EMG of gastrocnemius (MG) was 27.4 % higher when using arms compared to No Arms – condition. This could be explained by the change of COM position. When using arms as "wings" while perturbation, it changes the COM more anteriorly and more activation is needed from the posterior ankle segment muscles to correct the unbalanced position of the body back to straight. To support the COM changes, the SOL was showing significantly higher activation levels in EMG_{300} (150 – 300 ms) but not in EMG_{50}. COM is a variable which influences to the COP changes. While quiet stance the net COP lies somewhere between the two feet and varies when a weight of a person is changing directions. (Winter 1995.) In the present study, No Arms –condition decreased the COP movement (figure 18) and increased the TA activity (figure 11). Results indicated that the EMG activity of TA was somewhat (n.s.) higher when COP AP was lower. Winter (1995) reported similar findings earlier by stating that increased dorsiflexion (TA) activity moves COP posteriorly.

However, the unperturbed balance requires a solid use of hip segment too (Matjacic et al. 2001). Although this study revealed no significant difference in hip angle (6 % increase while arms), ES muscle which takes part of back extension was more active in No Arms –condition. Due to large standard deviation in No Arms –condition the difference, however, was not statistically significant. Smaller range of the motion of the hip angle in No Arms –condition could be explained by the very active use of back extensors (ES) which helped to hold the hip straighter. Other interesting explanation could be more rapid correction to the natural upright position when COM was moving. Loram et al. (2005) stated that COM lags 100 to 300 ms behind the muscle activity so rapid correction is needed. To support the trunk more rapidly, subjects also used the RA muscle (antagonist). Tokuno et al. (2013) have shown that, the abdominal muscles acts first followed by posterior trunk muscles while balance perturbation. Results of this study support that theory. The hypothesis of this study was that the hip strategy involves free use of arms, and when arms are crossed (No Arms –condition) the recovery from the perturbation involves more trunk muscle activity. The EMG of the RA muscle was almost 55 % higher in No Arms –condition ($p = 0.043$) and the ES on average two time higher in No Arms –condition with remarkable individual variations. However, it should be noticed that even though a huge variation in ES results, it is not significant based on the statistics which might be caused my low amount of subjects. When analyzing the EMG of the RA in three different time windows, the RA was significantly more active ($p < 0.05$) in EMG_{150} (50 – 150 ms) and in EMG_{300} (150 – 300 ms) (Figure 12) in No Arms –condition compared to Arms-condition. Kagawa et al. (2011) recently reported that recovery from sudden slipping took about 200 ms. Therefore, this study suggested similar situation (of slipping) when using arms, by resulting higher EMG activation level of the RA in EMG_{300} ($p < 0.05$). Interesting was also to investigate RA activity in individual subjects, since during the measurements, it was observed that one subject had much greater EMG amplitudes than the others. This was subject number four who was falling backward all the time so that the safety person needed to catch her. This subject's EMG results of the RA were also showing abnormal levels as seen in results figures 13 and 14. This could have influence to the results. Nevertheless, the comparison between Arms and No Arms –conditions in RA followed the trend.

When a human is a having a feeling of falling or slipping it is quite natural to try to grasp something or take a step and correct the unperturbed balance. Corbeil et al. (2013) observed reach-to-grasp strategy in their study, where they reported that the forward arm movements evoked during posterior falling motion (slipping) with a counterweight strategy. In this study, there was no handles or rails, but a subject was instructed take a step if needed, however it was preferred not to take the step. The results suggested a significantly higher activation level of the EMG of AD muscle ($p < 0.05$) with Arms-condition especially 50 to 150 ms after the perturbation ($p < 0.05$). It has been reported that arms voluntary activation occurs between 80 to 150 ms (McIlroy & Maki 1995). However, it is notable to realize that in No Arms –condition the freedom of the arm movement (ROM) has been taken away which can be one explanation to the higher EMG level in Arms-condition. According to the results, the PD muscle was showing opposite behavior. The PD was more active (n.s.) in No Arms –condition. This indicates strategy changes in deltoid muscle when comparing these two conditions.

Factors like reaction time, pre-activation of the muscles, fatigue and learning could have a small influence to the results. Skotte et al. (2004) concluded that the reaction time for sudden load to trunk muscle was faster after three trials, which is a good indicator of our learning strategy in the body. The learning factor was taken into consideration in this study by performing two to three pre-trials before the actual trial run. For the future studies it would be worth of re-think the protocol, perhaps it could be smaller and be more focused on three or four conditions instead of eight. The measurements for the MVC should have been done in more maximal way. In this study MVCs were done with a limitation because there was no power bench in the lab for the real maximal voluntary contraction.

Cross correlation analysis was used to assess the relationship between the two conditions, but due to small group (N=5) the statistics were not appropriate. For the discussion there were some signs of slight correlations between the velocity of the shoulder (ML direction, positive correlation) and COP as well as elbow angle (ML, negative correlation) and COP when using arms. But since the small N, this could not be done. For the future studies, this should be taken into consideration.

The aim of this study was to examine upper body strategy during random perturbation. The first plan was to focus on especially how arms and deltoid muscle involve to the control of balance and if the strategy changes when the treadmill translation direction changed from forward to backward. However, due to some technical problems related to triggering and data synchronization, only one direction of the translation was analysed in the results and discussion. For future studies, it would be interesting to analyse if the direction of the perturbation really did influence to the balance control –strategy when using or not using arms.

In conclusion the upper body muscle activation is higher when not using arms compared to the situation where arms are normally in use during forward perturbation. Because of the higher activation level needed from the upper body muscles (RA, ES and PD) during perturbation when no arms was used, people should take care of the adequate muscle strength. For example if you are slipping and holding something in your arms, you are not able to use your arms normally. That was the case in this study with No Arms –condition. Based on the findings in this study, it is important to work with the basic core muscles (RA and ES) to be able to handle sudden unbalance situations to avoid for example falling down.

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