The control of equilibrium in bimanual, whole-body lifting tasks

a biomechanical approach

Dianne Commissaris

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The work presented in this thesis was carried out at the Amsterdam *Spine* Unit,
in the Institute of Fundamental and Clinical Human Movement Sciences,
Faculty of Human Movement Sciences,
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ISBN 90-9010567-0
NUGI 821
Subject headings: control of equilibrium / biomechanics / whole-body lifting tasks

Printer: PrintPartners Ipskamp B.V. Enschede.
Cover design: Dianne Commissaris.

VRIJE UNIVERSITEIT

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ACADEMISCH PROEFSCHRIFT

ter verkrijging van de graad van doctor aan
de Vrije Universiteit te Amsterdam,
op gezag van de rector magnificus
prof.dr. T. Sminia,
in het openbaar te verdedigen
ten overstaan van de promotiecommissie
van de faculteit der bewegingswetenschappen
op donderdag 4 september 1997 om 13.45 uur
in het hoofdgebouw van de universiteit,
De Boelelaan 1105

door

Dimphena Adriana Cornelia Maria Commissaris

geboren te Rucphen

Promotor: prof.dr.ir. G.J. van Ingen Schenau
Copromotor: dr. H.M. Toussaint

if the human brain was so simple
that we could understand it,
we would be so simple
that we couldn't

n.n.

Voor Maurice en
mijn ouders

Parts of this thesis are derived from the following published or submitted papers:

Commissaris DACM, and Toussaint HM (1997)
Anticipatory postural adjustments in a bimanual, whole body lifting task with an object of known weight. *Human Movement Science* **16**: 407-431.

Commissaris DACM, and Toussaint HM (1997)
Load knowledge affects low-back loading and control of balance in lifting tasks. *Ergonomics* **40**: 559-575.

Commissaris DACM, Toussaint HM, and Hirschfeld H (submitted)
The coordination of movement and equilibrium in a bimanual, whole-body lifting task: I. A synergy between upper body and lower segments controls equilibrium. *Experimental Brain Research*

Commissaris DACM, and Toussaint HM (submitted)
The coordination of movement and equilibrium in a bimanual, whole-body lifting task: II. A key role for the ankle muscles. *Experimental Brain Research*

Commissaris DACM, Toussaint HM, and Hirschfeld H (submitted)
Is regulation of the anterior-posterior centre of mass position the leading principle in the organization of postural adjustments in a bimanual, whole-body lifting task? *Experimental Brain Research*

Contents

Chapter 1

General introduction

Introduction

Skilful performance of many human movements is not possible without adequate postural control. This statement is not only applicable to the performance of complex gymnastic exercises, such as making a somersault on the beam, but also to the performance of (what an observer would classify as) rather simple daily tasks. Adequate postural control is indispensable for a successful execution of all sorts of manipulative tasks performed in stance (Massion 1992), such as opening a door, receiving a tray full of cups, operating an engine, handing a toy to one's parent or washing a patient in bed. Moreover, postural control and its development play a crucial part in the development of successful, autonomous actions in children (Reed 1989). Usually, humans are not aware of the fact that they actively control the posture of a certain body segment or the equilibrium of the whole body during task execution. The task is, almost automatically, performed in a dexterous and successful manner. However, smooth and skilful performance of motor acts is beyond the reach of a large group of people, because they lack, among others, adequate postural control. Children with brain damage, for example, may develop only a limited repertoire of basic motor skills and may, depending on the nature and location of the damage, be never able to sit or walk independently or to manipulate objects. Adults can lose postural mechanisms that belonged to their repertoire for many years, due to an accident, an illness or the process of aging.

Skilful performance of many human movements is not possible without adequate postural control, because the execution of a manipulative task in stance often entails a disturbance of a given posture and/or of the whole-body's equilibrium (Massion 1992; 1994). A *disturbance of posture* occurs when the muscle forces that realize the intended or primary movement cause reactive torques which, due to the multi-linked structure of the body, affect the angular motions at joints that are not directly involved in the primary movement (Massion 1992; 1994). A fast flexion of the extended elbow joint, for example, will cause a torque into extension at the wrist joint when the forearm is supinated (Latash et al. 1995b). Following Cordo and Nashner (1982), the primary movement is termed a *focal* movement in this thesis. Since the position and orientation of body segments such as the head, trunk or arms serves as a reference frame for perception and action with respect to the external world (Berthoz and Pozzo 1994; Massion 1994; Savelsbergh and Kamp 1993), maintenance of the position and orientation of those segments against disturbances is essential for smooth and successful performance of the focal movement. To accomplish this, additional (i.e. non-focal) muscles should be activated whose resulting torques stabilize the segments that are not directly involved in, but involuntarily affected by the focal movement. These additional muscles are classified as *postural* muscles (Cordo and Nashner 1982).

A *disturbance of equilibrium* occurs when the focal movement involves a horizontal[1] displacement of the centre of mass (CoM) of the whole body with respect to the base of support (Gahery and Massion 1981; Massion 1992), as occurs when bending the upper body forward (Crenna et al. 1987; Oddsson and Thorstensson 1986) and when raising the arms in front of the body (Aruin and Latash 1995a; Friedly et al. 1984). To maintain equilibrium, that is to keep the CoM inside the base of support (in static conditions like upright stance) or over the future base of support (in dynamic conditions like walking), additional (*postural*) muscles should be activated to displace the CoM of body segments that are not involved in the focal movement in a direction that's opposite to the direction of the undesirable CoM shift caused by the focal movement. The postural muscles are located in the segments that support the body, most often the lower limbs (Bouisset and Zattara 1987; Cordo and Nashner 1982; Hirschfeld and Forssberg 1991). Thus, to achieve dexterous and successful performance of many daily activities, the disturbance of posture and/or equilibrium, that's inherent in the execution of the focal movement, has to be minimized by adequate postural mechanisms (Belen'kii et al. 1967; Massion 1992). The activity of postural muscles and the ensuing joint torques and segmental motions are usually classified as *anticipatory or associated postural adjustments*: actions that predictively counteract disturbances of posture and/or equilibrium that are associated with a voluntary movement (Bouisset and Zattara 1987; Cordo and Nashner 1982; Lee et al. 1987).

A common paradigm to investigate the control of equilibrium, the topic of this thesis, is the application of an external (i.e. inflicted by the experimenter) or internal (i.e. generated by the subject) perturbation to equilibrium (Aruin and Latash 1995b; Bouisset and Zattara 1987; Dietz et al. 1993; Nashner 1980). Equilibrium control counteracting the first type of disturbance is characterized by *reactions* to the external perturbation, while *anticipatory actions* are generated by the subject when the perturbation is self-inflicted and thus expected. The latter actions are more efficacious to counteract the equilibrium perturbation than the

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[1] *given the sagittal plane description of the lifting task in this thesis, the horizontal direction refers to the anterior-posterior direction*

first, since they are generated prior to the occurrence of the actual disturbance (Massion 1992). The control of equilibrium has been mainly investigated in *static tasks*, i.e. single or multi-joint *upper-body motions* in which the lower limbs are not involved in the focal movement, like pulling a handle (Cordo and Nashner 1982) or bending the upper body for- or backward (Oddsson 1988; Oddsson and Thorstensson 1986) in stance. The postural component predominates in this type of tasks. The results generally suggested that maintaining the horizontal position of the body CoM with respect to the base of support against expected or unexpected disturbances is the central rule that governs task organization (Massion 1992; 1994). In static tasks, it is feasible to identify muscles with a focal function, as contrasted to those with a postural role. The first ones are often located in the upper limbs or trunk, while the latter ones are situated in the lower limbs. This distinction in focal and postural actuators has, presumably, contributed to the view that two separate controllers are operative in the execution of voluntary movements, one regulating movement, the other equilibrium (Massion 1992; 1994).

Investigating the control of equilibrium in a static task is limited, though, in the sense that in daily life expected or unexpected equilibrium disturbances are often encountered in a *dynamic situation*, for example, while walking, rising from a seat, kneeling down or kicking a ball. In such *whole-body movements*, the legs are involved in both the focal movement and the postural adjustments, which means that these postural adjustments have to be integrated in the primary (focal) motion of the legs. Furthermore, the focal component predominates in many dynamic whole-body motions and a horizontal CoM displacement may be required to reach the task goal. Given the distinctions between static and dynamic tasks, the conclusions drawn with respect to the control of equilibrium in static, upper-body tasks may not be simply applicable to the control of equilibrium in dynamic, whole-body tasks. It may be questioned whether equilibrium control in such tasks is solely aimed at maintaining the horizontal position of the CoM with respect to the base of support. Would the control of a *position* suffice to maintain dynamic equilibrium and enable an undisturbed motion? Are additional principles or constraints perhaps operative to control equilibrium in a dynamic task? This topic has received little attention so far; just a few studies investigated the control of equilibrium during locomotion. Unexpected (Nashner 1980), as well as expected equilibrium perturbations (Hirschfeld and Forssberg 1991; Nashner and Forssberg 1986) have been applied to investigate how the normal locomotor pattern is adjusted to the altered equilibrium conditions.

Investigating the control of equilibrium in a bimanual, whole-body lifting task

The present thesis investigates the control of equilibrium in a bimanual, whole-body lifting task. This motor act comprises the forward bending of the trunk and the lowering of the body CoM to grasp an object with two hands, followed by the lifting of the object to waist or chest level (illustrated in Figure 1.1). For most humans, lifting and displacing large objects is a daily activity, performed in an occupational setting, during house keeping, nursing or as a leisure activity. In this dynamic task, the legs perform a part of the focal movement, i.e. lowering and raising the body, and they are assumed to be involved in the postural adjustments as well. It should be stressed that pronounced segmental rotations accomplish the whole-body motion. Therefore, task analysis ought to extend beyond the analysis of the translation of segments with respect to the base of support, as is common for static tasks (Crenna et al. 1987; Hirschfeld and Forssberg 1991). The biomechanical approach applied in this thesis will also analyse the task with respect to the rotation of segments and additional parameters, representing the (change in) whole-body rotation, will be introduced.

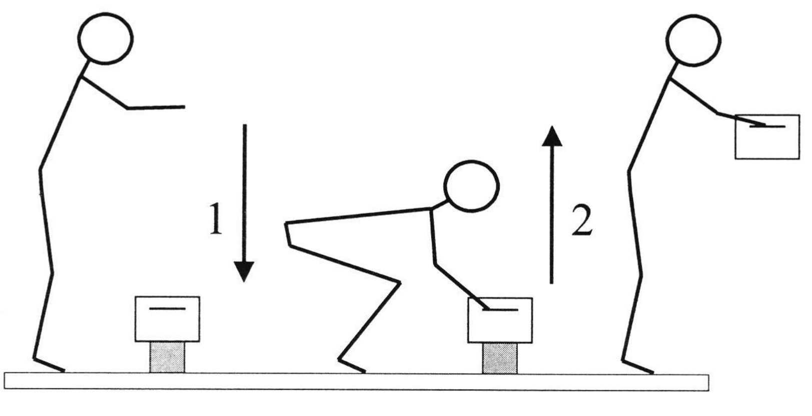


Figure 1.1
The investigated bimanual, whole-body lifting task. The unloaded downward movement (1) is followed by a loaded upward movement (2). The subject grasps and lifts the load in an ongoing motion.

The performance of a bimanual, whole-body lifting task is likely to entail a disturbance of equilibrium in two ways. In the first place, equilibrium maintenance is challenged by the forward and backward bending of the upper body, inducing, respectively, a considerable forward and backward displacement of the CoM of the upper body and, without adequate postural adjustments, of the total body. This aspect of the lifting task resembles the forward and backward upper-body movements performed in stance (Crenna et al. 1987; Massion et al. 1993; Oddsson 1990; Pedotti et al. 1989). The second challenge to equilibrium maintenance is inherent in the pick-up of an object in front of the body. The addition of an extra mass (anterior) to the body causes the CoM, of the combined body and load, to shift forward with respect to the base of support and, thus, presents the lifter with an expected perturbation to equilibrium. Furthermore, the forward shifting CoM induces a sharp decrease in the moment exerted by the ground reaction force with respect to the CoM. Given the fact that a rather large moment is required to promote the whole-body extension at this stage, the forward CoM shift could impede a smooth extending motion of the whole body if not adequately anticipated (chapter 2 will elucidate the relation between the moment exerted by the ground reaction force with respect to the CoM and the whole-body flexion and extension). This second type of equilibrium perturbation resembles the expected perturbation imposed upon a walking subject by instructing him or her to exert a force on a fixed handle (Hirschfeld and Forssberg 1991; 1992; Nashner and Forssberg 1986; Patla 1986).

The aims of the present thesis are:

(1) to examine whether postural adjustments counteract the expected equilibrium perturbations in a bimanual, whole-body lifting task. To make a clear distinction between the two types of equilibrium disturbance that are present in the lifting task, the term *associated postural adjustments* will be applied to classify the postural actions related to the forward and backward bending of the upper body and the term *anticipatory postural adjustments* will be reserved for the postural actions related to the addition of the load to the body.

(2) to determine the biomechanical aspects of the postural adjustments and of the interaction with the primary, focal motion. Following Winter (1995), who stated that "the total body kinetics are our window into the synergies of human movement", the global kinetics and kinematics of the lifting task are investigated.

(3) to examine the relation between the global kinematics and kinetics and the local joint torques and muscle activity patterns, in order to further characterize the basic strategies employed by the subjects to counteract the expected equilibrium perturbation.

(4) to determine whether and how the anticipatory postural adjustments are affected by the magnitude of the equilibrium perturbation (by varying the weight of the object) and by the dynamics of the primary motion (by varying the lifting technique).

(5) to determine whether and how the associated postural adjustments are affected by the base of support size.
(6) to explore the relation between the anticipatory postural adjustments and the loading of the lumbar spine, in view of the finding that whole-body lifting tasks are associated with low-back injuries in industry (Chaffin and Park 1973).
(7) to contribute to the formulation of a 'dynamic equilibrium' concept.

The seven issues listed above are addressed in the subsequent six chapters. Chapter 2 describes the biomechanical approach to the investigation of the control of movement and equilibrium in whole-body motions and defines the main parameters applied in this approach. Chapter 3 investigates the associated postural adjustments that are aimed at counteracting the disturbance of equilibrium that is inherent in the forward and backward bending of the upper body during the lifting task. Chapter 4 specifically examines if and how these postural adjustments are altered when the task is performed on a reduced base of support, augmenting the demands on equilibrium control. In these two chapters, the 'traditional' approach to the investigation of the control of equilibrium is applied, while in the next three chapters the 'new' biomechanical approach will be utilized. Chapter 5 explores the anticipatory postural adjustments that are aimed at counteracting the perturbation to equilibrium that is inherent in the pick-up of an object in front of the body. In addition, the influence of lifting technique on the anticipatory postural adjustments is investigated. Chapter 6 examines the effect of the presence or absence of correct load knowledge on the loading of the lumbar spine and the maintenance of equilibrium during the lifting task. In particular, the consequences of an erroneous (over)estimation of the load on the execution of the lifting task, the maintenance of equilibrium and the loading of the lumbar spine are addressed. Chapter 7 investigates whether the regulation of the anterior-posterior position of the CoM can be regarded as the leading principle in the organization of postural adjustments in the lifting task, as is traditionally proposed for upper-body tasks performed in stance.

The remainder of this first chapter contains a further review of relevant literature with respect to the control of posture and equilibrium, anticipatory parameter control and internal models to provide the broader theoretical background for the studies reported in this thesis.

Associated and anticipatory postural adjustments counteract an equilibrium disturbance

At the end of the 19th century, Babinski (1899) was one of the first to recognize the presence of active postural control during focal movements and to realize its significance. He noticed the inability of a patient with a cerebellar disfunction to bend the trunk and head backward during stance without falling backwards. Babinski attributed this to an absence of associated postural adjustments in the legs, because he observed that a healthy man flexed at the ankle and knee joints when performing the backward trunk motion, thus counteracting a large backward CoM shift. Later, Hess (1943) realized that performance of the focal movement itself was the source of the equilibrium perturbation. He suggested the presence of two components ("Kräftesysteme") in each motor act: the focal movement ("teleokinetisches Kräftesystem") and the equilibrium related postural adjustments ("ereismatisches Kräftesystem"). Hess proposed a central organization of both components of the motor act, a view that contrasted the general opinion at that time that posture was controlled by a simple peripheral feed-back system, based on the stretch reflex (Granit 1981). His vivid description of the two components may elucidate their respective functions:

> "Ein Obermann steht auf der Schulter eines Untermannes. Er springt ab auf ein vorgezeichnetes Ziel. Der Untermann fällt dabei rückwärts, der Obermann verfelht das Ziel (springt zu kurz). Der Obermann steht wieder auf der Schulter des Untermannes. Der Untermann wird aber durch einen Hintermann gesichert. Im Augenblick des Absprunges weicht der Untermann wieder aus. Er wird vom Hintermann stützend gehalten und hält so dem Rückstoss stand. Der Obermann gelangt zur ungestörten Entfaltung seiner propulsiven Kräfte und erreicht das Ziel" (p. C63, Hess 1943).

Belen'kii and co-workers (1967) were the first ones who actually recorded postural activity in leg muscles, together with focal activity in the shoulder muscle during the raising and lowering of the arm. They demonstrated the early activation of the postural muscles and concluded that "the anticipatory activation of the muscles as preparation for voluntary movement is connected with the need to maintain balance with the minimum expenditure of energy" (p. 161, Belen'kii et al. 1967). This classical study served as an example for many groups that investigated human movement and postural control. The groups of, for instance, Massion (e.g. Massion 1992), Gurfinkel (e.g. Gurfinkel 1973), Nashner (e.g. Nashner 1977) and more recently of Hirschfeld (e.g. Hirschfeld and Forssberg 1991), McIlroy (e.g. McIlroy and Maki 1993), Latash (e.g. Latash et al. 1995a) and Oddsson (e.g. Oddsson 1990) are well-known in this area.

The postural adjustments associated with a focal movement are called *anticipatory* because the onset of the change in postural muscle activity occurs *before* the onset of the equilibrium disturbance due to the focal movement (Massion 1992). Thus, a feed-forward postural control accompanies movement control, implying a central organization of both components of the motor act. Massion (1992) proposed to apply the term 'anticipatory' only in connection with postural adjustments that result from a voluntary movement, to emphasize that they ensue from an internal command and not from external feed-back. Following his proposition, the term 'anticipatory postural adjustments' should be reserved to those conditions in which (1) the postural movement has a direction that is opposite to the direction of the perturbation caused by the focal movement ("opposite sign", Bouisset and Zattara 1987), (2) the postural muscle activity disappears or is notably reduced in magnitude when extra support is provided to the whole body or to the postural segments (Cordo and Nashner 1982; Friedly et al. 1984; Nardone and Schieppati 1988) and (3) the postural muscle activity occurs simultaneously with or prior to, but never in reaction to the equilibrium perturbation (Massion 1992). Following the last point, the occurrence of an *anticipatory* activation of the postural muscles is intriguing, because, in theory, a *simultaneous* activation would suffice to counteract the perturbation associated with the focal movement (Latash et al. 1995b). Moreover, the motion of postural body segments is in itself a source of perturbation and must, therefore, occur only shortly before or simultaneously with the focal movement (Oddsson and Thorstensson 1986). Latash (1995b) considered two factors that may influence the 'choice' of the central nervous system (CNS) between anticipatory and simultaneous activation of postural muscles. In the first place, the postural muscle may have a long and compliant tendon, a characteristic that will delay the time between muscle activation and the generation of torque that is sufficiently large to counteract the perturbation expected from active generation of torque in another joint. In the second place, most of the postural muscle have a slower rate of torque generation than muscles involved in (fast) voluntary limb movements. Since this implies that a perfect compensation of the perturbing motion can in principle not be attained, the control system will apply anticipatory activation of postural muscles to minimize the expected perturbation (Latash et al. 1995b).

The studies that have been undertaken since the pioneer study of Belen'kii et al. (1967) have greatly enlarged the body of knowledge with respect to the control of human movement and the role of anticipatory postural adjustments in it. The prevailing view now emphasizes the flexibility of postural control and its adaptability to different contexts (Massion 1994). The older concepts of fixed postural synergies (Horak and Nashner 1986; Nashner 1977) appear no longer valid. At present, however, the exact location of postural networks or neural pathways responsible for the anticipatory postural adjustments is still unknown (Massion 1992). According to Massion's review (1992), the postural networks may be located at a rather low CNS level, i.e. the brain stem or the spinal cord, whereas higher levels (motor cortex and pyramidal tract) may be only involved in the acquisition of new postural patterns. The integration of movement control and postural control is suggested to be organized in either a hierarchical or a parallel mode (Massion 1992). In the first mode, the pathways controlling movement performance have collaterals acting on the postural networks, which implies that the onset of the anticipatory postural adjustments is time locked to the onset of the focal movement. In the latter mode, postural and movement control are organized rather independently in two parallel pathways, but the circuits involved in movement performance are inhibited until the postural adjustments have created a situation in which the perturbation to equilibrium due to the focal movement will be minimized. Both modes of coordination between postural and movement control imply that two separate controllers are operative, one regulating movement, the other equilibrium. Recently, Latash and colleagues (Aruin and Latash 1995b; Latash et al. 1995b) have questioned the strict dichotomy in the control of posture and movement. They suggested that postural adjustments may not be a mere addition to a voluntary motor command, but an inherent part of it. According to this view, the focal and postural commands are generated by one common controller (Aruin and Latash 1995b).

The studies and examples mentioned above demonstrate that the function of postural control is not merely one of stabilizing body segments against the force of gravity. Posture is more than a summation of all the anti-gravity stretch reflexes in the body, as Sherrington and Magnus thought in the beginning of this century (Reed 1989). Posture and movement are tightly integrated in daily activities (Reed 1989) and postural control is essential in any dynamic motor act, not only in static ones (Massion 1992).

Anticipatory parameter control relies on internal representations of the physical properties of the effector system and the object to be handled

As pointed out above, the execution of anticipatory postural adjustments must rely on feed-forward mechanisms, because the onset of the change in postural muscle activity occurs before the onset of the actual equilibrium disturbance associated with the focal movement (Massion 1992). Thus, the magnitude and direction of the future equilibrium disturbance should somehow be estimated and the anticipatory postural adjustments required to counteract this disturbance should be predicted on the basis of this estimation and taking into account the body dynamics and support conditions at the moment of the future equilibrium disturbance (Hirschfeld and Forssberg 1991). Recently, several authors have suggested that internal models of the physical properties of the body, the objects to be handled and task characteristics are applied in the control of anticipatory postural adjustments (Ghez et al. 1991; Lacquaniti et al. 1992). This idea emerged, among others, from the investigation of manipulative actions that involve impact forces, such as catching a falling ball with one hand (Lacquaniti et al. 1992) or dropping a ball from one hand into the other (Johansson and Westling 1988b). To prevent a disturbance of the posture of the hand and arm involved in catching, the impact force exerted by the ball on the hand and the resulting torques at the elbow, wrist and shoulder joints have to be counteracted by active torque generation at those joints. Indeed, when a ball was dropped from one hand into a receptacle held by the other, an anticipatory increase in grip force (applied by the fingers which held the receptacle) appeared about 150 ms prior to the impact and, simultaneously, the receptacle was lifted a little to meet the impact of the ball (Johansson and Westling 1988b). These anticipatory actions took into account the weight of the object, the weight of the receptacle, the height of the drop and the friction between hand and receptacle grip, implying a reliance on sensory-motor representations of relevant object and task properties. Lacquaniti et al. (1992) have shown that anticipatory muscle activity during catching a falling ball was directed at minimizing hand compliance (in world space) around impact. Given the dependence of hand compliance on both the patterns of muscle activity and the geometrical configuration of the arm, an internal model of limb geometry was suggested in the transformation of intended hand compliance into the appropriate muscle activity patterns (Lacquaniti et al. 1992).

Also, internal models or so-called sensory-motor memory representations (Gordon et al. 1993) of the physical properties of objects were proposed to govern the (focal) force generation in a manipulative task. Johansson and co-workers investigated the grasping and lifting of objects with a precision grip and demonstrated that the motor behaviour in this manipulative action depends largely on predictive rather than on servo-control mechanisms (Johansson 1991; Johansson and Cole 1992; Johansson and Edin 1993). The results of their studies support the common notion that healthy adults are able to tune their manipulative behaviour to the physical properties of an object, to the goal of the action and to environmental conditions. Moreover, they are able to adequately respond to small perturbations that occur during task execution, such as the sudden slip of a wet glass or moving liquid inside a cup. Noteworthy is that all this is possible before sensory feed-back has signalled the exact weight of an object, or the structure of its surface material, or the exact magnitude of the perturbing force. Johansson et al. (1993) termed the predictive motor behaviour *anticipatory parameter control*: the ability to parameterize the default motor commands in advance of the movement. One parameter that must be adapted to an object property is the vertical force exerted upward on the object. This force may only be slightly larger than the object's weight to prevent a large positional overshoot after lift-off. Explicit weight information, however, is first available after lift-off. Therefore, the weight is estimated on the basis of previous experience with that particular object (i.e. somato-sensory information is used) or on the basis of a size-weight transformation, under the assumption of a common density (i.e. visual or haptic information is used) (Gordon et al. 1991a; 1991b; 1991c). Next, the vertical force is adjusted to the estimated weight, meaning that the rate of increase in vertical force is reduced prior to lift-off of the object, such that the force only slightly exceeds the expected weight after lift-off (Forssberg et al. 1991; 1992; Gordon et al. 1993). While object weight was shown to determine the duration of the in-contact phase during precision grip lifting, the (perceived) fragility of an object was found to control the movement towards the object and in particular the closing of the fingers during that approach (Savelsbergh et al. 1996).

Anticipatory parameter control enables humans to perform fast, smooth and accurate (manipulative) motor behaviour despite the time delays associated with neural transmission (Rack 1981); before the relevant physical properties of an object are sensed or before the actual disturbance of posture and/or equilibrium has been detected, action can be taken on the basis of expectations and accurate predictions. The drawback of anticipatory parameter control, however, lies within a possible erroneous estimation of the object's physical properties and, consequently, an erroneous adjustment of the focal or postural muscle activity to those properties. In case the object's weight is overestimated, the vertical force will be too large at the expected moment of lift-off, yielding a jerky motion and a large positional overshoot (Johansson and Westling 1988a). On the other hand, underestimating the object's weight will lead to a failure to start the vertical movement at the expected moment of lift-off, because the vertical force is too small (Johansson and Westling 1988a). In both cases, corrective forces are triggered by respectively the presence of unexpected and absence of expected sensory information from tactile receptors in the glabrous skin on the palm of the hand (Johansson 1991; Johansson and Vallbo 1983; Johansson and Westling 1988a). Next to triggering corrective actions, the spatio-temporal pattern of sensory information in parallel afferent (mainly tactile) channels is applied to update the sensory-motor representations of the object's physical properties (Johansson and Cole 1992; Johansson and Edin 1993).

Chapter 2

A biomechanical analysis of dynamic multi-joint movements

Introduction

A two-dimensional biomechanical analysis is applied to study the control of equilibrium and the integration in the primary multi-joint movement of the postural actions that counteract the equilibrium disturbances associated with the performance of a bimanual, whole-body lifting task. In this analysis the lifter's motions, the underlying muscle activity patterns, the net leg joint torques and the external forces and moments of the external forces, all in the sagittal plane of motion, are collectively examined to infer basic mechanical and/or neural principles about and constraints in the control of equilibrium. At the heart of the biomechanical analysis is the idea that the lifter controls the external ground reaction force during task execution, an idea that is based on the work of Ingen Schenau and colleagues (Ingen Schenau et al. 1992; 1995a; Jacobs and Ingen Schenau 1992a). These authors demonstrated, in a variety of tasks, that the actor controls the direction of an external force by regulating torques at leg joints (Bobbert and Ingen Schenau 1988; Doorenbosch and Ingen Schenau 1995; Ingen Schenau et al. 1992; Jacobs and Ingen Schenau 1992a). Moreover, the observed leg muscle activity patterns could be completely understood in the light of the requirement to control the direction of the external force. A similar approach was applied by Toussaint and co-workers in studying coordinative aspects of bimanual, whole-body lifting tasks (Toussaint et al. 1992; 1995).

The heart of the biomechanical analysis, applied by Toussaint and colleagues and employed in this thesis, is the interdependence of local and global mechanics. Local muscle contractions induce (by means of torque generation) changes in the segmental linear and angular momenta, which, summed over all segments, equal the changes in the linear momenta of the body's CoM and the angular momentum of the whole body with respect to the CoM. Considering on a global level the whole body (including the load after pick-up) as a single free body, the external ground reaction force determines the linear momenta of the body's CoM (given the force of gravity) and the moment effect of the ground reaction force about the CoM (the external moment, Toussaint et al. 1995) determines the angular momentum of the whole body with respect to the CoM. Thus, the global parameters magnitude, direction and point of application of the ground reaction force determine, taking into account the position of the body's CoM, the dynamic behaviour of the whole body in terms of linear and angular momenta. The global parameters magnitude, direction and point of application of the ground reaction force are the net result of local muscle contractions which generate torques at multiple joints. Recent studies have suggested that the force exerted on the ground is not merely a passive consequence of the acceleration and deceleration of body mass, but that it is a high-level parameter that is controlled by the CNS (Ingen Schenau et al. 1992; Macpherson 1988a; 1988b). It was found that the direction of the external force is controlled by an adequate distribution of torques across the hip and knee joints (Ingen Schenau et al. 1992; Jacobs and Macpherson 1996; Macpherson 1988a; Toussaint et al. 1992) and that the point of application of the ground reaction force is controlled by activity of muscles crossing the ankle joint (Crenna and Frigo 1991; Okada and Fujiwara 1984).

The biomechanical parameters

The global biomechanical parameters applied in chapters 5 to 7 are the magnitude and direction or the vertical and horizontal component of the ground reaction force, the point of application of the ground reaction force (i.e. the centre of foot pressure, CoP), the horizontal and vertical momenta of the system CoM (i.e. body or body and object), the external moment exerted by the ground reaction force about the system CoM and the angular momentum of the whole system about its CoM. The local parameters are the net torques at the ankle, knee, hip and lumbo-sacral joints and electromyographic (EMG) recordings of eight superficial leg muscles. The biomechanical analysis is restricted to the sagittal plane of motion. The caudo-cranial and dorsal-ventral directions and joint torques having an extending effect are defined positive. A two-dimensional linked segment model of the body, described by Looze et al. (1992), is used to estimate the mass, CoM, moment of inertia, position, (angular) velocity and (angular) acceleration of eight segments (feet, lower legs, upper legs, pelvis, trunk/head, upper arms, forearms, hands/load) and to estimate the net torques at the joints, using an inverse dynamics analysis (Elftman 1939).

The *ground reaction force magnitude* is computed from the magnitude of the horizontal and vertical components of the ground reaction force vector (Figure 2.1). The *ground reaction force direction* is defined as the angle of the force vector with respect to the ground.

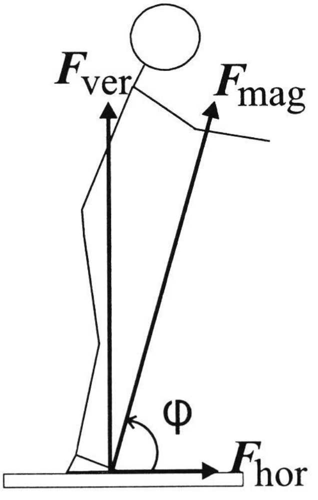


Figure 2.1
Definition of ground reaction force magnitude (F_{mag}) and direction (φ).

The vertical component of the ground reaction force (F_{ver}) is related to the rate of change in the vertical momentum of the system CoM ($\dot{p}_{ver}$) according to:

$$F_{ver} - F_{gravity} = \dot{p}_{ver} \qquad (2.1)$$

with the *vertical momentum* of the system CoM, p_{ver}, calculated from the vertical momenta of the individual segments:

$$p_{ver} = \sum_{j=1}^{8} (m_j \cdot v_{verj}) \qquad (2.2)$$

where m_j and v_{verj} represent respectively the mass and vertical velocity of the jth segment. Likewise, the horizontal component of the ground reaction force (F_{hor}) is related to the rate of change in the horizontal momentum of the system CoM ($\dot{p}_{hor}$) according to:

$$F_{hor} = \dot{p}_{hor} \qquad (2.3)$$

with the *horizontal momentum* of the system CoM, p_{hor}, calculated from the horizontal momenta of the individual segments:

$$p_{hor} = \sum_{j=1}^{8} (m_j \cdot v_{horj}) \qquad (2.4)$$

where m_j and v_{horj} represent respectively the mass and horizontal velocity of the jth segment. The *horizontal positions of the CoM and CoP* are expressed as a percentage of the base of support according to:

$$y_{rel} = \frac{y - y_{heel}}{y_{toe} - y_{heel}} * 100\% \qquad (2.5)$$

where y_{rel} and y represent respectively the relative and absolute horizontal position of the CoM and CoP and y_{heel} and y_{toe} are the horizontal positions of a marker on respectively the heel and the distal end of the most prominent toe.

The *external moment* (M_{ext}) is calculated according to:

$$M_{ext} = \mathbf{a} \times F_g \qquad (2.6)$$

where **a** is the vector from the CoM of the body and load to the CoP and F_g is the ground reaction force vector. Figure 2.2 illustrates the external moment.

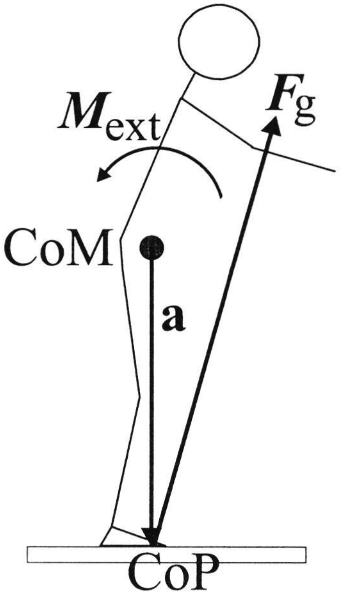


Figure 2.2
Moment effect of the ground reaction force (M_{ext}) about the centre of mass.

The external moment equals the rate of change in the angular momentum of the entire system (Toussaint et al. 1995), analogous to the relation between $\boldsymbol{F}_g$ and linear momenta:

$$\boldsymbol{M}_{ext} = \dot{\boldsymbol{L}} \qquad (2.7)$$

with the *angular momentum* of the system, $\boldsymbol{L}$, calculated from the angular momenta of the individual segments according to Greenwood (1965):

$$\boldsymbol{L}=\sum_{j=1}^{8}(\boldsymbol{I}_j\cdot\boldsymbol{\omega}_j+(m_j\boldsymbol{r}_j\times\boldsymbol{v}_j)) \qquad (2.8)$$

where I_j is the moment of inertia of the jth segment relative to the segmental CoM, ω_j the angular velocity of the segment, m_j the segment mass, $\boldsymbol{r}_j$ the position vector from the segment CoM to the system CoM and $\boldsymbol{v}_j$ the velocity of the segment CoM relative to the system CoM.

Thus, the angular momentum of the whole body is composed of the angular momentum of each segment with respect to its own CoM and the angular momentum of each segment relative to the total CoM. Figure 2.3 illustrates the contribution of the angular momentum of the trunk/head and hands/load segments to the total angular momentum. The contribution of the trunk/head in the whole-body angular momentum is considerable, due to the large moment of inertia and a substantial rotational velocity (thus, $I\bullet\omega$ is large). The contribution of the hands/load segment is also considerable, though, because a rather large mass (for example 20% of the total mass) is positioned at a large distance from the total CoM (thus, $m\bullet\boldsymbol{r}\times\boldsymbol{v}$ is large).

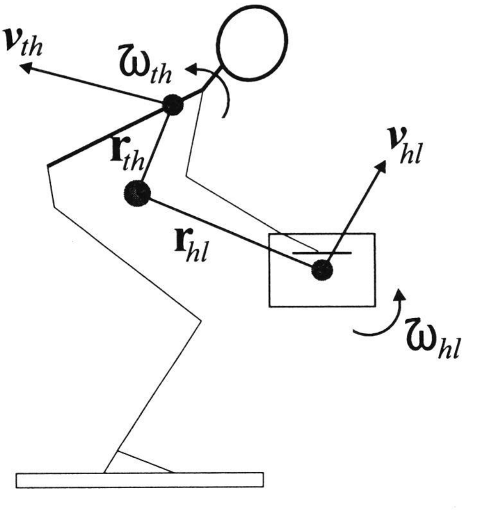


Figure 2.3
Contribution of the angular momentum of the trunk/head (th) and hands/load (hl) segments to the whole-body angular momentum.

Chapter 3

The coordination of movement and equilibrium in a bimanual, whole-body lifting task: I. A synergy between upper body and lower segments controls equilibrium

Abstract

The kinematics of a bimanual, whole-body lifting task were described and analysed to study the organization of equilibrium control in a multi-joint movement, in which the legs are involved in both the primary movement and the associated postural adjustments that control equilibrium. Nine healthy male subjects grasped and lifted a 6.7 kg box, in an ongoing down- and upward motion, from 0.14 m above the floor to chest height. In this task, equilibrium maintenance is challenged by the forward and backward bending of the upper body and by the pick-up of the load that implies a forward shift of the centre of mass (CoM) (of body and load) with respect to the base of support. A kinematic synergy was revealed, that comprised a simultaneous forward displacement of the head and shoulder, backward displacement of the pelvis and forward displacement of the knee during the downward phase and resulted in a minimal forward displacement of the body CoM. It was concluded that the multi-joint movement in a bimanual, whole-body lifting task is coordinated such that the position of the CoM is regulated with respect to the base of support. Although changes in the ankle, knee and hip joint angles served both a focal and a postural role, the postural adjustments related to load pick-up appeared primarily in the ankle joint angle and not in the knee and hip joint angles.

Introduction

Skilful performance of a motor act requires that the execution of the goal directed movement is accompanied with adequate postural adjustments which minimize the disturbance to posture or equilibrium caused by the primary movement (Belen'kii et al. 1967; Massion 1992). Voluntary upper trunk movements in standing humans, for example, were found to be accompanied with simultaneous hip and knee movements in the opposite direction (Crenna et al. 1987; Massion et al. 1993; Oddsson 1990; Pedotti et al. 1989). The associated movements of the legs were classified as postural adjustments, directed at counteracting the displacement of the body's centre of mass (CoM) with respect to the base of support, that is inherent in the primary movement of the upper body (Crenna et al. 1987; Oddsson 1988; Oddsson and Thorstensson 1986; Pedotti et al. 1989). Maintenance of the horizontal position of the CoM relative to the base of support against external and internal disturbances is generally considered crucial in the control of equilibrium (Belen'kii et al. 1967; Massion 1992; 1994). The coordination between the primary upper trunk movement and the associated postural adjustments seems to be organized at a central level, since the simultaneous or prior onset of activation of postural relative to prime mover muscles rules out peripheral feed-back processes (Crenna et al. 1987; Oddsson and Thorstensson 1987; Pedotti et al. 1989). Babinski (1899) was one of the first to recognize the presence of postural control during goal directed trunk movements and to realize its significance. He noticed the inability of a patient with a cerebellar disfunction to bend the upper body backward during upright stance without falling backwards. Babinski attributed this to an absence of associated postural adjustments in the legs, because he observed that a healthy man flexed at the ankle and knee joints during performance of the backward trunk motion, thus counteracting a large backward CoM shift. The combination of a voluntary upper-body motion in one direction and associated lower limb movements in the opposite direction was called a "synergy" by Babinski (1899) and some authors later extended this to "axial synergy" (Alexandrov et al. 1994; Crenna et al. 1987; Massion et al. 1993; Pedotti et al. 1989).

The goal directed movement and the associated postural adjustments are rather easily identified in the kinematics of a voluntary upper trunk movement: the forward/backward trunk motion can be classified as goal directed, the opposite directed hip and knee movements as postural adjustments (Crenna et al. 1987; Oddsson 1990; Pedotti et al. 1989). The same holds for various other tasks, in which the primary movement is restricted to rotations in one or two joints and in which the supporting segments are only involved in the postural adjustments, not in the goal directed movement such as arm movements in stance (Belen'kii et al. 1967; Bouisset and Zattara 1987; Cordo and Nashner 1982; Friedly et al. 1984; Lee et al. 1987). However, a bimanual, whole-body lifting task (i.e. lowering the body and bending the upper body forward to grasp an object with two hands and subsequently lift it to waist or chest level) is a multi-joint movement, in which the goal directed action involves changes in both upper trunk inclination and hip, knee and ankle joint angles. Given the necessity to prevent a (too) large horizontal CoM displacement, the forward/backward upper-body motion has to be accompanied with an opposite directed displacement of other segments, more than likely the legs. This would imply that the supporting segments are involved in performance of both the primary movement and the associated postural adjustments. The question rises as to how equilibrium control is organized in this special case of voluntary upper-body motion. Is the commonly observed synergy between upper body and lower limbs still present? If not, how is the horizontal CoM position with respect to the base of support then controlled? Furthermore, in view of the major role of the ankle joint muscles in the control of equilibrium (chapter 5; Burleigh et al. 1994; Crenna and Frigo 1991; Crenna et al. 1987; Massion et al. 1993; Oddsson 1989; Oddsson and Thorstensson 1987; Pedotti et al. 1989), it might be questioned whether the associated postural adjustments are controlled by changes in all three leg joint angles or primarily by changes in the ankle joint angle.

The aim of the present study was to investigate the organization of equilibrium control in a multi-joint movement, in which the legs are involved in both the goal directed action and the associated postural adjustments. The main questions were (1) how the position of the CoM is maintained with respect to the base of support and to what extent and (2) whether the ankle has a more pronounced role in control of the associated postural adjustments than the knee and hip. To that end, we studied a bimanual, whole-body lifting task, in which subjects grasped and lifted a 6.7 kg box, in an ongoing down- and upward motion, from 0.14 m above the floor to chest height. In this task, equilibrium maintenance is not only challenged by the for- and backward bending of the upper body, but also by the pick-up of the load that implies a forward shift of the CoM (of body and load) with respect to the base of support. To separate both equilibrium-disturbing effects, we also studied for- and backward bending of the upper body without load pick-up, i.e. imitating the lifting motion. The present chapter investigates the kinematics of the lifting task performed under normal circumstances and chapter 4 will examine both kinematics and leg muscle activity patterns when the task is performed in a condition in which the base of support was restricted to 0.092 m in the anterior-posterior direction.

Methods

Subjects

Nine healthy male subjects (mean age 23.7 ± 2.6 (standard deviation -SD-) years, body height 1.80 ± 0.04 m, body mass 71.3 ± 9.0 kg, shoelength 0.277 ± 0.010 m) participated in both experiments, after they had given written informed consent and after approval of the Faculty's ethical committee. None of the subjects reported a history of low-back disorders or other motor impairments.

General procedure and experimental protocol

The subjects were instructed to grasp a 6.7 kg box with two hands in the middle of an ongoing downward-upward movement and to lift it in a symmetric motion to chest height. To investigate the above mentioned first challenge to equilibrium maintenance (i.e. upper-body movement) without the second challenge (i.e. load pick-up), the subjects performed several unloaded down- and upward movements before they actually grasped and lifted the box. While subjects were moving up and down in a free-chosen rhythm, one of the authors counted down to the moment of box grasping ("grasp"), starting at the beginning of the penultimate downward phase. The data acquisition started about 0.5 s prior to that moment and ended about 1 s after the box had reached the intended position. Thus, successively, one completely unloaded movement cycle and one movement cycle with an unloaded downward and a loaded upward phase were recorded (illustrated in Figure 3.1). No specific instructions were given regarding lifting technique, but all subjects performed the task by bending and extending both the legs and the lumbar spine. The box was placed in front of the subject at a standardized position; both the horizontal distance between the front of the feet and the back of the box handles and the vertical distance between the base of support and the bottom of the handles was 15% of the individual body height (0.270 ± 0.006 m). Subjects executed this task eight times in a row, separated by a 20-s pause. They were instructed to apply the same speed of movement in each trial, to keep the heels on the ground during the whole task and to restrict their movements to the sagittal plane. Before the experiment, the subjects performed about ten practice trials to familiarize themselves with the task and the load.

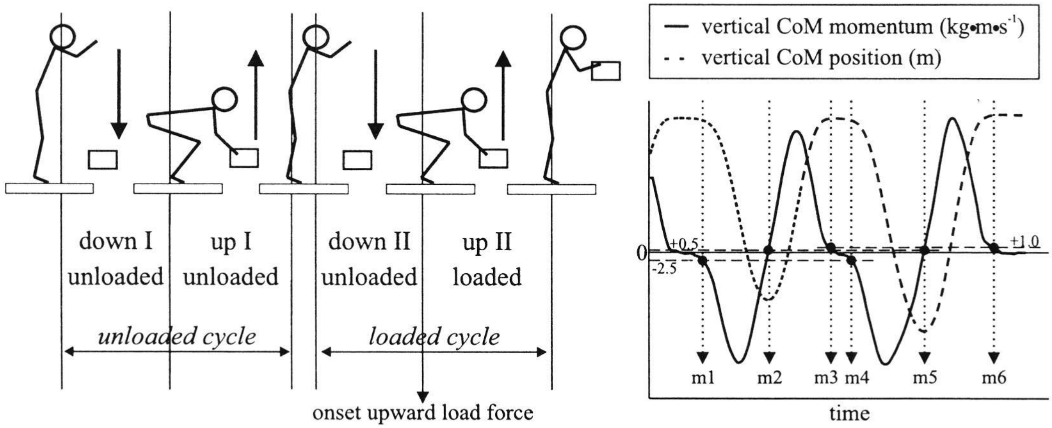


Figure 3.1
Schematic outline of the bimanual, whole-body lifting task and definition of the four movement phases. The vertical body centre of mass (CoM) position and momentum are time traces of a typical example. The position time trace was re-scaled (enlarged) and shifted (downwards) to match the (unmodified) scaling of the momentum time trace. Abbreviations m1 to m6 are explained in the text.

Apparatus and data acquisition

A special box, equipped with ten strain gauge force transducers was used in this study. The box consisted of a solid metal frame to which a (replaceable) synthetic cover (width {left-right} 0.400 m, height 0.200 m, depth {anterior-posterior} 0.265 m) was firmly attached. Two handles were positioned on both sides of the box, 0.131 m above the bottom and 0.046 m away from the side. The weight of the empty box (frame, handles and cover) was 6.7 kg. On each side of the frame, close to the handles, three transducers recorded forces in three orthogonal directions. In each handle, two additional transducers recorded the sagittal plane squeeze forces (grip forces). The analog force signals were amplified, low-pass filtered (30 Hz, 4th order), sampled (60 Hz, 12 bits) and stored synchronously to the recorded movements by a motion registration system (see below). The force signals were low-pass filtered off-line with a digital filter (6 Hz, 2nd order Butterworth, zero phase lag). In the present study, only the vertical force (the left and right forces added) was used to determine the moment at which subjects started to exert an upward force on the box (i.e. the first positive increase in force) and to determine the moment of lift-off of the box (i.e. vertical force exceeded the box's weight).

A 3-D semi-automatic video-based motion registration system (VICON™, Oxford Metrics Ltd.) recorded the positions of fourteen light-reflecting markers (ø 25 mm), with four cameras, at a sample rate of 60 Hz. Eleven markers were attached to the subject's right body side, to indicate the location of the fifth metatarsophalangeal joint (on the shoe), the ankle joint (distal part of lateral malleolus), the knee joint (lateral epicondyle), the hip joint (greater trochanter), the lumbo-sacral joint (L5-S1) (as in Looze et al. 1992), the spinous process of the first thoracic vertebra (Th1), the head (caput mandibula), the lateral border of the acromion, the elbow joint (lateral epicondyle), the wrist joint (ulnar styloid) and the hand (third metacarpophalangeal joint). Three markers were placed on the right side of the box. The coordinates of the acromion marker were used to determine the position of the shoulder joint. The raw sagittal plane coordinates of the markers were low-pass filtered off-line with a digital filter (6 Hz, 2nd order Butterworth, zero phase lag). The coordinates of the joint marker positions defined nine body segments: the feet, lower legs, upper legs, pelvis, trunk, neck and lower part of head, upper arms, forearms and hands(/load). Note that we use the term 'trunk' to define the segment from L5-S1 to Th1, and not 'upper trunk'.

Data analysis and statistics

Per subject, the mass and the CoM position of each segment, except the trunk, were calculated according to Plagenhoef et al. (1983) and Looze et al. (1992). The mass and CoM location of the feet were recalculated to include the mass and CoM location of the shoes. The sagittal plane box CoM is situated in the middle of the box handle and its coordinates were calculated relative to the coordinates of the three box markers. The mass and CoM location of the hands were adapted in the time period between the moment at which subjects started to exert an upward force on the box and the moment of box lift-off to include the mass and location of the CoM of the box. The coordinates of the markers on Th1 and L5-S1 were used to determine the position of the trunk CoM during the movement according to an optimization procedure, which improved the estimated horizontal trajectory of the body CoM (Kingma et al. 1995). The whole-body CoM was calculated from the mass and location of the segmental CoMs. To separately analyse the motions of upper body versus lower limbs, the upper-body CoM and lower-limbs CoM were calculated from the mass and location of the segmental CoMs of, respectively, the trunk, head, upper arms, forearms and hands(/load) and the feet, lower legs, upper legs and pelvis. The angle of each segment was calculated relative to the horizontal. To quantify the goal directed trunk motion, the angular inclination of the trunk (L5-S1 to Th1) was calculated relative to the vertical (Figure 3.2), derived from the definition of angular trunk (L3-C7) inclination with respect to the vertical of Oddsson and Thorstensson (1986). Joint angles (enclosed angle between two adjacent segments) were calculated in ankle, knee and hip (Figure 3.2). Numerical differentiation (Lanczos 5-point differentiation filter) of the time histories of the angles and marker positions yielded (angular) velocities. For translations, the caudo-cranial and dorso-ventral directions were defined positive and for rotations an extending joint motion. In case an angular displacement could not be classified as flexing or extending (e.g. change in trunk inclination), the counter-clockwise direction was defined positive.

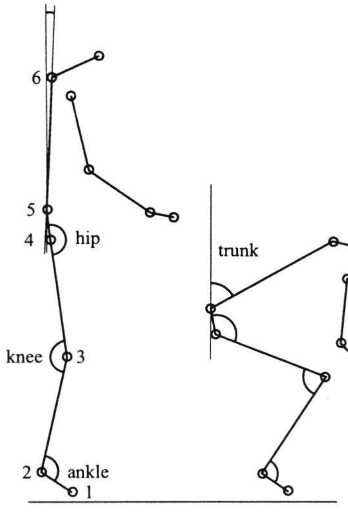


Figure 3.2
Definition of the angle of trunk inclination and the angles at the ankle, knee and hip joint. The open circles represent the end points of the nine links. The numbers correspond to markers at the (1) fifth metatarso-phalangeal joint, (2) ankle joint, (3) knee joint, (4) hip joint, (5) lumbo-sacral joint, (6) first thoracic vertebra.

Marker trajectories, CoM position and displacement and angular changes were investigated during the second (loaded) movement cycle and compared with those of the first (unloaded) cycle. A movement cycle comprises a down- and subsequent upward phase. Figure 3.1 illustrates the method we applied to define the onset and end of each down- and upward phase. Six instants (m1 to m6), which coincided with the extremes in vertical CoM position, were marked in the time trace of the vertical body CoM momentum (right panel). The three criteria used to mark these instants were determined by visual inspection of several trials of each subject. The highest CoM position, defining the onset of the first and second downward phase, is marked by m1 and m4 (criterium: first sample at which the vertical momentum was lower than -2.5 kg•m•s^{-1}). The lowest CoM position around the end of a downward/onset of an upward phase is marked by m2 and m5 (criterium: first sample at which the vertical momentum exceeded +0.5 kg•m•s^{-1}). The end of down II/onset of up II was set at the onset of the upward load force on the box. The end of down I/onset of up I was obtained by subtracting from m2, the time difference between the onset of the upward load force and m5. The highest CoM position, defining the end of the first and second upward phase, is marked by m3 and m6 (criterium: first sample at which the vertical momentum was lower than +1.0 kg•m•s^{-1}).

To obtain average results, the mean of the parameter value was calculated for eight trials per subject and next the nine mean-subject values were averaged. A Principal Components Analysis (PCA) was applied on the time traces of the trunk inclination and hip, knee and ankle joint angles to investigate the covariation in the angular changes. This analysis was, except for the input data, similar to the PCA applied by Alexandrov et al. (1994) and Mah et al. (1994). The PCA was performed on the angles of the unloaded and loaded movement cycle separately, for each trial and each subject. The significance of differences between average results was checked with a paired samples *t*-test. A Kolmogorov-Smirnov Goodness of Fit Test was applied to test whether the output parameters of the PCA were a sample from the normal distribution. In each test, an alpha level of 0.05 was employed to determine statistical significance.

Results

The downward phase was characterized by a for- and downward motion of the Th1, head and upper limb markers (upper body), a down- and backward displacement of the L5-S1 and hip markers (pelvis) and a down- and forward movement of the knee joint marker. In the upward phase, the upper body moved up- and backward again, as did the knee joint marker and the moved pelvis up- and forward. The body CoM showed a marked vertical, but only a small horizontal displacement (Figure 3.3). The average horizontal displacement of the joint, head and Th1 markers and of the body CoM during the loaded cycle (Table 3.1) confirms the picture presented in Figure 3.3: the hip and L5-S1 joint markers showed a net horizontal displacement that was opposite to the displacement of the other markers in both movement phases and the net horizontal displacement of the body CoM remained small. The maximal horizontal CoM excursion (i.e. the range of motion within one phase) during the loaded cycle was also small: average values were 0.025 (±0.014) m during the downward and 0.046 (±0.013) m during the upward phase. The higher magnitude of the latter value was caused by the forward shift of the CoM (on average 0.038 (±0.006) m, visible in the right picture of Figure 3.3 for the typical example) due to the addition of the load mass to the body at the onset of the upward phase. During task execution, the horizontal velocity of the body CoM remained small: the maximum forward velocity was on average 0.065 (±0.029) $m \bullet s^{-1}$ and the maximum backward velocity 0.080 (±0.024) $m \bullet s^{-1}$ during the loaded cycle.

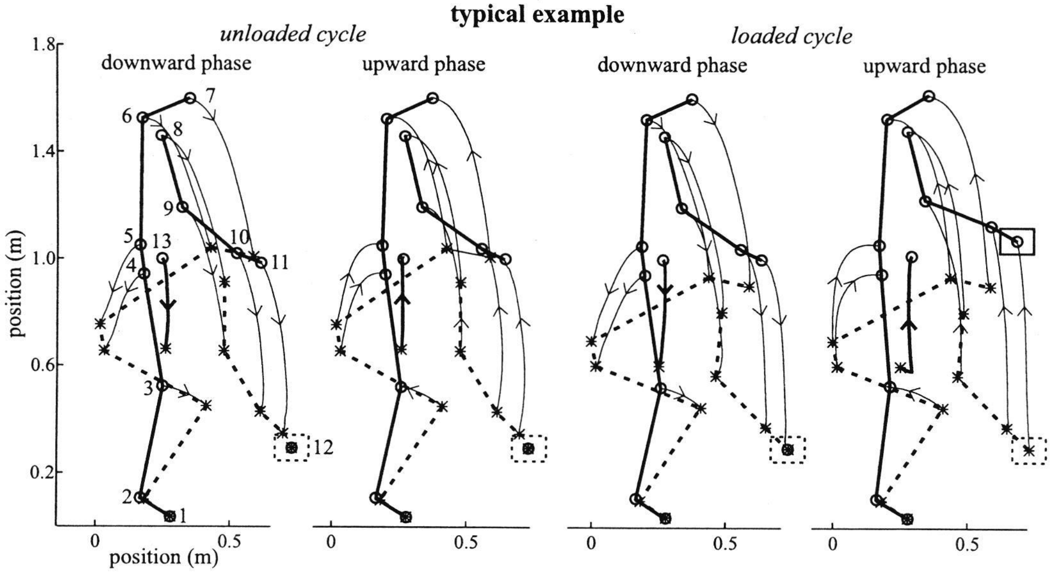


Figure 3.3
Position of markers and body centre of mass (CoM) at the onset and end of the downward phase (open circles and asterisks respectively) and at the onset and end of the upward phase (asterisks and open circles respectively) of the unloaded and subsequent loaded movement cycle, presented for a typical example. The linked segment model, joining the markers, is indicated by thick solid lines at the onset of a downward and end of an upward phase and by thick dashed lines at the end of a downward and onset of an upward phase. The trajectory and direction of displacement of the markers and CoM during the down- and upward movement are indicated with thinner lines and arrows respectively. The numbers correspond to markers at the (1) fifth metatarsophalangeal joint, (2) ankle joint, (3) knee joint, (4) hip joint, (5) lumbo-sacral joint, (6) first thoracic vertebra, (7) head, (8) shoulder joint, (9) elbow joint, (10) wrist joint, (11) hand, (12) box CoM and (13) body CoM (including the load after pick-up).

Table 3.1
Net horizontal displacement of the joint markers, of the markers on the head and the first thoracic vertebra (Th1) and of the body centre of mass (CoM) (including the load after pick-up) during the downward and during the upward phase of the loaded movement cycle. Mean values ± 1 SD (*n=9*) are presented. L5-S1: lumbo-sacral joint. Marker- and CoM trajectories are illustrated in Figure 3.3.

Marker	horizontal displacement (m)			
	downward phase		upward phase	
ankle	+0.015	(0.004)	-0.016	(0.004)
knee	+0.173	(0.046)	-0.201	(0.042)
hip	-0.161	(0.029)	+0.143	(0.033)
L5-S1	-0.164	(0.024)	+0.149	(0.028)
Th1	+0.231	(0.042)	-0.268	(0.052)
head	+0.220	(0.039)	-0.267	(0.048)
shoulder	+0.216	(0.031)	-0.233	(0.035)
elbow	+0.222	(0.089)	-0.186	(0.097)
wrist	+0.243	(0.165)	-0.129	(0.098)
body CoM	+0.013	(0.021)	+0.006	(0.027)

+ and - indicate a for- and backward displacement respectively.

At the end of the upward phase, all joints and segments were positioned close to the location obtained at the onset of the downward phase (Figure 3.3). The marker trajectories appeared similar in the down- v. upward phase, only of opposite direction, and in the unloaded v. loaded cycle, except for the hip and L5-S1 joint marker trajectories. These markers showed a simultaneous down- and backward movement during the downward phase, whereas they successively moved upward and then forward during the upward phase. The delayed start of the forward hip and L5-S1 motion was more pronounced in the loaded cycle: the average delays between the onset of the forward hip and L5-S1 motions and the onset of the upward phase were, respectively, 78 (±102) and 79 (±77) ms in the unloaded cycle v. 251 (±88) and 241 (±77) ms in the loaded cycle (hip: t=5.93, p<0.001, L5-S1: t=4.74, p=0.001). Furthermore, the duration of the loaded upward phase was significantly longer than that of the unloaded upward phase: 1027 (±106) ms v. 884 (±45) ms (t=4.82, p=0.001). It should be noted that the pelvis segment orientation in space hardly changed during task execution for the typical example (Figure 3.3). Other subjects did show a larger change in segment orientation during task performance, but the inter-individual variation was considerable. The average total change in pelvis segment angle relative to the vertical during the two movement cycles recorded was 10.8 ± 4.2°.

Despite the pronounced net horizontal displacement of all joint markers, except the ankle, and of the head and Th1 marker, the net horizontal displacement of the upper-body CoM and the lower-segments CoM remained rather small during the loaded movement cycle (compare the CoM displacements in Table 3.2 with the marker displacements in Table 3.1). During the downward phase, the lower-segments CoM shifted a little backward, whereas the upper-body CoM moved forward. As a result, the total-body CoM moved forward only 0.013 m (Table 3.2). Picking up the load mass in the beginning of the upward lifting phase caused a for- and downward shift of both the upper and total-body CoM, but after this initial shift both CoMs showed a more or less vertical upward trajectory (see Figure 3.4, which illustrates the trajectories of the CoMs of the upper body, lower segments and total body for the typical example). Note that the lower-segments CoM of the typical example first moved upward and even a little backward and then forward (Figure 3.4), a trajectory that resembles the hip and L5-S1 joint marker trajectories shown in the loaded upward phase of Figure 3.3. The net horizontal displacement of the three CoMs was on average less than 1 cm during the upward phase (Table 3.2). Thus, the small net horizontal displacement of the body CoM was accompanied by minor net horizontal displacements of both the upper-body CoM and the lower-segments CoM.

Table 3.2
Horizontal position at the onset of the downward phase and net horizontal displacement during the downward and during the upward phase of the loaded movement cycle for the centre of mass of the lower segments (below the lumbo-sacral (L5-S1) joint marker), the upper body (above L5-S1) including the load after pick-up and the total body including the load after pick-up. Mean values ± 1 SD ($n=9$) are presented. The centre of mass trajectories are illustrated in Figure 3.4.

Centre of mass	position (m)		horizontal displacement (m)			
	at onset down		downward phase		upward phase	
lower segments	0.191	(0.025)	-0.017	(0.027)	<-0.001	(0.027)
upper segments	0.241	(0.024)	+0.036	(0.025)	+0.004	(0.034)
total body	0.220	(0.023)	+0.013	(0.021)	+0.006	(0.027)

+ and - indicate a for- and backward displacement respectively.

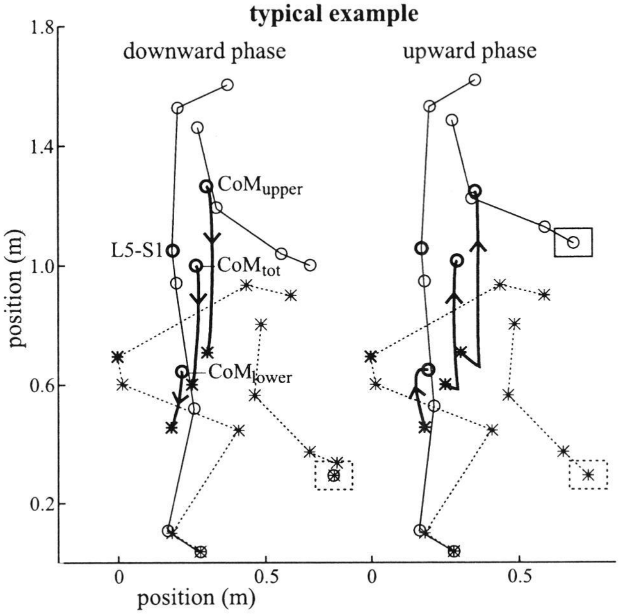


Figure 3.4
Position of the centre of mass of the lower segments (CoM_{lower}), of the upper body (CoM_{upper}, including the load after pick-up), of the total body (CoM_{tot}, including the load after pick-up) and of the lumbo-sacral (L5-S1) joint marker (separating lower segments and upper body) at the onset and end of the downward phase (open circles and asterisks respectively, left panel) and at the onset and end of the upward phase (asterisks and open circles respectively, right panel) of the loaded movement cycle, presented for the typical example. The linked segment model, joining the markers, is indicated by thin solid lines at the onset of a downward and end of an upward phase and by thin dashed lines at the end of a downward and onset of an upward phase. The trajectory and direction of displacement of the CoMs during the down- and upward movement are indicated with thicker lines and arrows respectively.

To further analyse the goal directed movement and associated postural adjustments, we reduced the complex multi-joint lifting movement to the motion of the upper-body CoM with respect to the L5-S1 joint marker and the motion of L5-S1 with respect to the ankle joint marker (Figure 3.5). We left the forward movement of the knee joint marker with respect to the ankle out of consideration, because this did not affect the main point we wish to make here. Furthermore, we chose the 'rotation centre' between both body parts at the L5-S1 joint, since the orientation of the pelvis segment in space hardly changed during task execution. We regarded the angle of the upper-body link relative to the vertical as a measure of the goal directed upper-body movement, whereas we conceived the angle of the lower-segments link as a measure of the associated postural adjustments. As all upper-body segments moved for-and downward relative to the L5-S1 joint marker during the downward phase of the loaded cycle, the clockwise rotation of the upper link with respect to L5-S1 was large (typical example in Figure 3.5, average results in Table 3.3). Hence, the upper-body CoM shifted forward relative to L5-S1. At the same time, the lower link rotated counter-clockwise with respect to the ankle joint marker (Figure 3.5, Table 3.3), yielding a backward displacement of the L5-S1 joint marker relative to the ankle. The combined forward shift of the upper-body CoM with respect to L5-S1 and the backward motion of this joint marker with respect to the base of support yielded the small horizontal displacement of the upper-body CoM shown in Table 3.2.

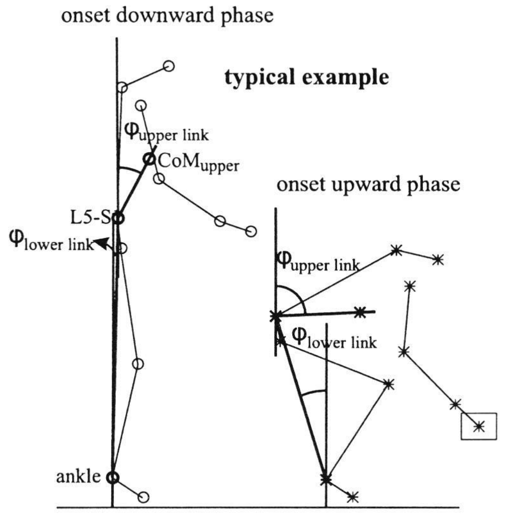


Figure 3.5
A simplification of the complex multi-joint lifting movement into a two-joint motion. The upper-body link (lumbo-sacral (L5-S1) joint marker to upper-body centre of mass (CoM_{upper})) rotates with respect to the L5-S1 joint, the lower-segments link (ankle to L5-S1) rotates with respect to the ankle joint. Upper link angle ($\phi_{upper\ link}$: L5-S1 to CoM_{upper}, relative to vertical) and lower link angle ($\phi_{lower\ link}$: ankle to L5-S1, relative to vertical) are shown at the onset of the downward (left panel) and upward phase (right panel) of the loaded movement cycle for the typical example.

Table 3.3
Angle at the onset of the downward phase and net angular displacement during the downward and during the upward phase of the loaded movement cycle for the upper-body link (L5-S1 to CoM_{upper}, relative to vertical) and lower-segments link (ankle to L5-S1, relative to vertical). Mean values ± 1 SD (n=9) are presented. The two body links are defined and illustrated in Figure 3.5.

Link	angle (deg)		angular displacement (deg)			
	at onset down		downward phase		upward phase	
upper body	-23.1	(3.1)	-61.6	(7.7)	+49.1	(7.9)
lower segments	-0.7	(1.1)	+16.1	(3.2)	-15.3	(3.3)

+ and - angular displacement indicate a counter-clockwise and clockwise rotation respectively.

The effectiveness of the counter-clockwise rotation of the lower link and related backward L5-S1 joint marker displacement, i.e. the associated postural adjustments that accompanied the goal directed clockwise rotation of the upper link, is evaluated in Figure 3.6. The positions of the joint markers, the head, the upper-body CoM, lower-segments CoM and total-body CoM at the onset of the loaded upward phase were estimated in case of absence of postural adjustments (right panel) and compared with the positions obtained with postural adjustments (left panel). The upper-body and lower-segments CoM would have been positioned respectively 0.175 (±0.024) and 0.098 (±0.004) m anterior to the locations actually obtained. As a result, the total-body CoM would not have shifted only 0.013 m forward during the downward phase, but 0.155 m.

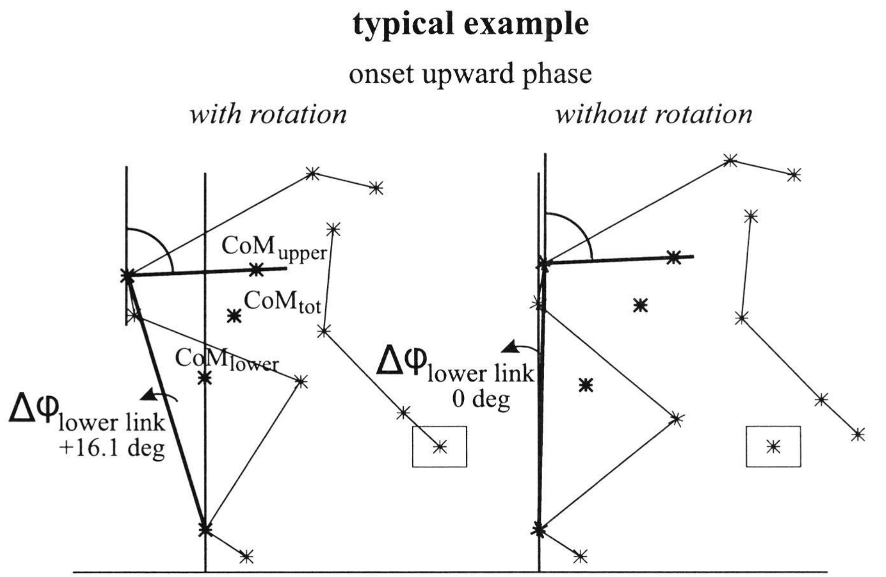


Figure 3.6
Position of the centre of mass of the lower segments (CoM_{lower}), the upper body (CoM_{upper}) and the total body (CoM_{tot}) at the onset of the upward phase of the loaded movement cycle, with (left panel) and without (right panel) the counter-clockwise rotation of the lower link. The two-link model is defined and illustrated in Figure 3.5. $\Delta\phi_{lower\ link}$ is the rotation of the lower link (L5-S1 to ankle), relative to the vertical during the downward phase.

Both the down- and upward movement were accomplished by rotation of segments, causing considerable changes in trunk inclination and in joint angles (see for instance Figure 3.2).

In the downward phase of the loaded movement cycle, all joints flexed and the trunk bent forward (Table 3.4). The angular changes in the upward phase were opposite in direction and about equal in magnitude, compared with those of the downward phase (Table 3.4). The four time traces of the ankle, knee and hip joint angles and of the trunk inclination during one movement cycle showed a remarkable resemblance (compare the pattern of the time traces between the four panels of Figure 3.7). Moreover, the time traces over two movement cycles and eight consecutive trials of the same subject were invariant, suggesting that this subject was able to perform the lifting task in a stereotyped manner (compare pattern and amplitude of the sixteen time traces within each panel of Figure 3.7). Only the ankle angle displayed a different time course in the loaded v. unloaded cycle: the onset of ankle extension relative to the onset of the upward phase occurred somewhat earlier in the loaded movement cycle (for this subject: 15 (±36) ms in the loaded v. 138 (±126) ms in the unloaded cycle, t=2.50, p=0.041). However, the average results did not confirm the typical example results in this case: the onset of ankle extension occurred earlier, but not significantly earlier in the loaded cycle (-21 (±55) ms in the loaded v. +9 (±74) ms in the unloaded cycle, t=1.83, p=0.104).

Table 3.4
Angle at the onset of the downward phase and net angular displacement during the downward and during the upward phase of the loaded movement cycle in the ankle, knee and hip joints and of the trunk inclination with respect to the vertical. Mean values ± 1 SD (n=9) are presented. Angles are illustrated in Figure 3.2.

Joint	angle (deg)		angular displacement (deg)			
	at onset down		downward phase		upward phase	
ankle	108.4	(2.1)	-20.3	(5.5)	+23.8	(5.1)
knee	163.0	(5.5)	-83.7	(12.6)	+88.8	(11.8)
hip	190.9	(10.1)	-57.1	(6.4)	+57.6	(6.4)
trunk inclination	3.4	(3.7)	-56.9	(9.2)	+59.4	(10.9)

+ and - angular displacement indicate, respectively, a joint extension/counter-clockwise change in trunk inclination and joint flexion/clockwise change in trunk inclination.

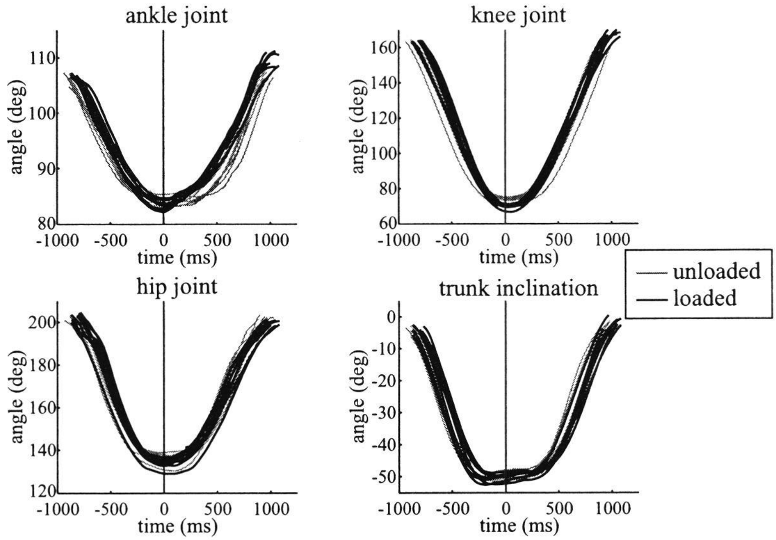


Figure 3.7
Time traces of the angles in ankle, knee and hip joints and of the trunk inclination relative to the vertical during the unloaded and loaded movement cycle of eight consecutive trials of one subject. Each time trace is presented from the onset of the downward to the end of the upward phase. The curves are synchronized in time to the onset of the upward force on the box (*t*=0) and the corresponding moment in time in the unloaded movement cycle (see data analysis section).

The PCA, applied on the time traces of the trunk inclination and the hip, knee and ankle joint angles during each movement cycle for each subject and every trial, showed that the first principal component (PC1) accounted for 99.1% (median value) of the variance in the ensemble of all four angles for the unloaded movement cycle and 98.2% of the variance for the loaded cycle (Table 3.5). PC2 accounted for only 0.65% of the variance in all four angles in the unloaded cycle and 1.5% in the loaded one (median values are shown, because some output parameters of the PCA were not a sample from the normal distribution.). The percentage of variance in each individual angle explained by PC1 was lowest for the ankle joint; 99.4% in the unloaded and 98.3% in the loaded cycle. Noteworthy is that the percentage of variance, both in the ensemble of the four angles and in each individual angle, that could be explained by PC1 was lower for the loaded than for the unloaded movement cycle.

Table 3.5
Results of the Principal Components Analysis (PCA), applied on the time traces of the trunk inclination and hip, knee and ankle joint angles during the unloaded and loaded movement cycle of the lifting task. The PCA was performed for each subject and each trial separately, yielding seventy-two cases per movement cycle. Median values and range of values (minimum-maximum) are presented for the percentage of variance in the ensemble of the four angles that could be accounted for by each of the four Principal Components (PC1 to PC4) and for the percentage of variance in the individual angles that could be accounted for by the first Principal Component (PC1). Angles are illustrated in Figure 3.2.

	unloaded cycle		loaded cycle	
	median	range	median	range
% explained variance by:				
PC1	99.10	97.30 - 99.80	98.20	92.10 - 99.50
PC2	0.65	0.10 - 2.60	1.50	0.40 - 7.80
PC3	0.10	0.00 - 0.50	0.20	0.00 - 0.60
PC4	0.00	0.00 - 0.00	0.00	0.00 - 0.20
% explained variance by PC1 in:				
ankle	99.40	97.20 - 99.99	98.30	93.20 - 99.80
knee	99.90	99.50 - 99.98	99.80	99.30 - 99.96
hip	99.80	98.60 - 99.95	99.60	98.30 - 99.90
trunk inclination	99.40	97.60 - 99.90	98.70	91.30 - 99.70

Discussion

Investigating the control of movement and equilibrium in a complex multi-joint movement

The control of movement and equilibrium has been studied mainly in tasks in which one part of the body performed the goal directed or "focal" (Cordo and Nashner 1982) movement and another part was involved in the associated postural adjustments. This paradigm presumably provided the experimental evidence for the often proposed separate pathways for control of movement and equilibrium (Massion 1992; 1994). However, the investigation of complex multi-joint movements, such as a bimanual, whole-body lifting task, will not readily provide evidence for a separate control of movement and equilibrium, because several segments will serve both a focal and a postural function. Recently, Latash and colleagues (Aruin and Latash 1995b; Latash et al. 1995b) have proposed that a common central command may control both the goal directed movement and the associated postural adjustments in a motor act. They postulated that "the classification of the joints into 'focal' and 'postural' may be done by experimenters but not by the central nervous system" (p. 299, Aruin and Latash 1995b). The scheme of one common controller for movement and equilibrium implies that the associated postural adjustments are "not an addition to a 'voluntary motor command', but an inherent component of it" (p. 299, Aruin and Latash 1995b). This control scheme might be applicable to investigate the control of multi-joint movements, in which the legs have both a focal and a postural role.

Equilibrium control in a bimanual, whole-body lifting task

The bimanual, whole-body lifting task investigated in the present study contains a voluntary bending of the upper body, but, contrary to the upper-body motions studied before, the supporting segments, the legs, are also involved in the primary movement. Given the large mass of the upper body and its elevated position in relation to a rather small support surface, any horizontal displacement of the upper body implies an internal disturbance of the reference value maintained in equilibrium control, i.e. the horizontal position of the body CoM with respect to the base of support (Massion 1994). It was argued that control of the position of the trunk in standing man is crucial in the regulation of equilibrium (Gurfinkel et al. 1981; Oddsson 1989). Thus, the forward/backward trunk motion in the bimanual lifting task is expected to be accompanied with an opposite directed movement of the leg segments. The first question raised in this study was to what extent the horizontal body CoM position is maintained with respect to the base of support and what patterns of inter-joint coordination accomplish that. The second question was whether the ankle has a more pronounced postural role than the knee and hip. The first question was addressed by investigating equilibrium maintenance during the forward and backward bending of the upper body in both the unloaded and loaded movement cycle, the second one by comparing the kinematic patterns of the loaded and unloaded cycle, thus, investigating the way in which equilibrium was maintained when a mass was added to the body.

A synergy between upper body and lower limbs regulates the position of the body CoM with respect to the base of support

The forward displacement of the head, shoulder and Th1 markers was accompanied with a backward displacement of the pelvis (quantified by hip and L5-S1 displacement) during the downward phase. Hence, the horizontal displacement of the upper-body CoM (the segments above L5-S1) remained small. This pattern of coordination is comparable to the (axial) synergy described for forward trunk bending by Crenna et al. (1987) and Oddsson and Thorstensson (1986) and, as proposed by those authors, it appears to be the kinematic outcome of an integrated control of movement and equilibrium. The flexion at the ankle, knee and hip joints, which controlled part of the primary downward movement of the whole body, was coordinated such that the horizontal displacement of the lower-body CoM (the segments below L5-S1) was also small. The combination of joint motions in the upper body and lower segments resulted in a minimal forward shift of the whole-body CoM during the downward phase. It seems justified to conclude that the horizontal position of the CoM was indeed *regulated* with respect to the base of support, as the net horizontal CoM displacement during each movement phase remained small, even in the loaded upward phase which started with a forward CoM shift of 0.038 m. However, it is not justified to state that the horizontal CoM position was *maintained* with respect to the base of support, since horizontal CoM excursions (i.e. range of motion within one phase) up to 0.076 m were observed during the movement. To our opinion, the combined forward displacement of the head and shoulder, backward displacement of the pelvis and forward displacement of the knee in this lifting task can be classified as a synergy between upper body and lower segments, although a backward knee motion was observed during the previously characterized synergy in forward trunk bending (Crenna et al. 1987). The forward, instead of backward, knee motion was a consequence of the applied lifting technique and did not substantially affect the backward pelvis displacement, because the ankle flexion (underlying the forward knee motion) occurred simultaneously with knee flexion. If subjects had used a back-lift technique (i.e. no angular change at the knee joint) to lift the load, the 'classic' synergy of a horizontal trunk motion in one direction and a knee and hip joint displacement in the opposite direction would presumably have been observed.

The effectiveness of this complex inter-joint coordination in regulating the CoM position is stressed by our estimation that without associated postural adjustments in the legs the body CoM would have shifted on average 0.155 m forward during the downward phase, and not just 0.013 m. In this procedure, we recalculated the positions of the joint markers and of the CoMs of upper body, lower segments and total body and did not simulate the lifting movement. Consequently, the estimation did not take into account the possible ranges of natural joint motion (the typical example's ankle joint angle appears unnaturally small in Figure 3.6), nor the position of the hands with respect to the load (the typical example fails to grasp the load), nor the mechanical coupling of trunk and pelvis, through which the forward trunk movement would have caused a passive backward pelvis motion. Thus, we assume that we have overestimated the forward CoM shift. Ramos and Stark (1990) estimated the contribution of active associated postural adjustments during simulated forward trunk bending and concluded that the forward CoM displacement would be about 0.09 m in absence of active adjustments, while it was only 0.01 m in their presence. Surprisingly, their simulation predicted that the body would fall backwards in the end without associated postural adjustments, due to the backward perturbation of the lower segments caused by forward trunk bending (Ramos and Stark 1990).

A special role for the ankle joint in the control of associated postural adjustments?
Picking up the box and lifting it to a certain height imposed extra demands on equilibrium control in the loaded movement cycle, compared with the unloaded one. The addition of the load mass to the body at the onset of the upward phase caused a quick forward shift of the CoM (of body and load) of on average 0.038 (±0.006) m. This brought the average projection of the CoM onto the ground only 0.079 (±0.018) m behind the front margin of the base of support, which is even in front of the fifth metatarsophalangeal joint. Previous studies on equilibrium control in a bimanual, whole-body lifting task have revealed that this forward CoM shift is counteracted by an anticipatory increase in the backward CoM velocity *before* picking up the load (chapter 5; Toussaint et al. in press 1997a). The present study showed that an additional measure was taken *after* load pick-up: the forward motion of the hip and L5-S1 joint markers was delayed until completion of a major part of the upward movement (right picture of Figure 3.3). The delay of the forward hip motion yielded an adjusted trajectory of the lower-segments CoM in the loaded upward phase: first back- and upward and only forward after completion of the upward displacement (right panel of Figure 3.4). Without delay of the forward hip joint marker motion, the total-body CoM would have shifted even further towards the front margin of the base of support, implying an increased risk of loosing balance. Only a precise coordination of the angular changes in knee and ankle joints could accomplish an upward hip movement without a simultaneous forward motion. The delay of the forward hip joint marker motion appeared to be achieved by an earlier onset of ankle extension at the start of the loaded upward phase, compared with the unloaded upward phase, while the onset of knee extension did not change (upper panels of Figure 3.7). In the unloaded movement cycle, the ankle and knee extension started on average 9 (±74) and 9 (±33) ms after the onset of the upward phase. In the loaded cycle, ankle extension

started on average 21 (±55) ms before and knee extension 13 (±40) ms after the onset of the upward phase. The difference in onset of extension was indeed not significant for the knee joint (t=1.06, p=0.322), but it was also not for the ankle joint (t=1.83, p=0.104).

The angular changes at the ankle joint showed the same global pattern as the angular changes in trunk inclination and at the knee and hip joints. This is evident for the typical example presented in Figure 3.7 and confirmed by the results of the Principal Components Analysis: the first Principal Component (PC1) explained respectively 99.4%, 99.9%, 99.8% and 99.4% of the variance in the time traces of the ankle joint angle, knee joint angle, hip joint angle and trunk inclination during the unloaded movement cycle. Keeping in mind that the change in trunk inclination represents the focal movement only and that the angular changes at the leg joints represent both the focal movement and the associated postural adjustments, the strong covariation of the changes in the four angles may be the result of an integrated control of focal movement and postural adjustments. It should be remarked, though, that the U-shaped time trace of the four angles underlies the largest part of the variation (ànd covariation) in the time course of all four angles. As compared with the knee and hip, the angular changes at the ankle joint did not seem to fulfil a special postural role during the unloaded cycle. The angular changes at the ankle joint, however, were more influenced by the extra equilibrium constraint posed by picking up the load than the angular changes at the knee and hip joints. The percentage of variance in the ankle joint angle explained by PC1 was reduced from 99.4% in the unloaded movement cycle to 98.3% in the loaded one, whereas the reductions in the percentage of variance in the knee and hip joint angles explained by PC1 were only 0.1% and 0.2% respectively. Thus, the extra postural adjustment related to load pick-up was to a large extent manifested in the ankle joint motion and not in the angular changes of other joints with a possible postural function. The postural role of the ankle joint did not clearly emerge, though, because the U-shaped time profile of the ankle joint angle remained and still dominated the amount of variation in time course, much more than the small adjustments in time trace related to load pick-up did.

Thus, investigation of the way in which equilibrium was maintained when a load was picked up and lifted, provided some evidence that angular changes at the ankle joint may have a special role in the execution of associated postural adjustments. We found a tendency of an earlier onset of ankle extension at the start of the loaded upward phase, compared with the unloaded upward phase, while the onset of knee extension did not change. These events presumably underlay the delay of the forward hip motion at the start of the loaded upward phase. This delay, in turn, prevented that the total-body CoM, undergoing a considerable shift forward at load pick-up, moved even further towards the front margin of the base of support during the initial part of the upward phase. Since we feel that our data do not provide unequivocal evidence in favour of a special role for the angular changes at the ankle joint in the coordination between movement and equilibrium in a bimanual, whole-body lifting task, we will further assess this role in chapter 4. We will augment the demands on equilibrium control during lifting by letting the same subjects perform the same lifting task on a base of support that is reduced to 0.092 m in the anterior-posterior direction. In that way we attempt to create a situation in which the hypothesized role of the ankle joint in the control of associated postural adjustments will emerge more profoundly.

Acknowledgements

The authors would like to thank Idsart Kingma, Michiel de Looze, Jaap van Dieën and Gerrit Jan van Ingen Schenau for reviewing the manuscript.

Chapter 4

The coordination of movement and equilibrium in a bimanual, whole-body lifting task: II. A key role for the ankle muscles

Abstract

This chapter investigated the role of ankle muscles in the coordination between the goal directed movement and associated postural adjustments in a bimanual, whole-body lifting task. In this multi-joint movement, the leg muscles have both a focal and a postural function. Nine healthy male subjects grasped and lifted a 6.7 kg box, in an ongoing down- and upward motion, from 0.14 m above the support surface to chest height. The task was executed on a base of support of normal size and on a base of support that was reduced to 0.092 m in the anterior-posterior direction (a beam). It was expected that the latter condition would augment the demands on equilibrium control and, thus, create a situation in which the hypothesized role of the angular changes at the ankle joint in the control of equilibrium would emerge. Kinematics of the task and electromyographic (EMG) patterns of eight leg muscles were recorded. The kinematic synergy that was revealed during lifting on the normal base of support was in essence maintained during task performance on the beam, while adjustments in range of motion of the ankle joint resulted in an adequate and effective adaptation of the trajectory of the centre of mass (CoM) with respect to the reduced base of support. Less flexion in the ankle yielded a more backward positioned pelvis and, thus, the CoM projection remained close to the middle of the beam throughout the movement. This was accomplished by a substantial change in the EMG pattern of muscles crossing the ankle joint, whereas muscles crossing the knee and hip joints did not show such major changes in EMG pattern. It was concluded that, although the angular changes at the ankle, knee and hip joints served both a focal and a postural role in this task, the associated postural adjustments were primarily regulated at the ankle joint.

Introduction

In the preceding chapter, the kinematics of a bimanual, whole-body lifting task were described and analysed to study the organization of equilibrium control in a multi-joint movement, in which the legs are involved in both the primary movement and the associated postural adjustments that control equilibrium. In this task, equilibrium maintenance is not only challenged by the forward and backward bending of the upper body, but also by the pick-up of the load that implies a forward shift of the centre of mass (CoM) (of body and load) with respect to the base of support. To separate the equilibrium-disturbing effect of the upper-body movement on the one hand and of the addition of the load mass to the body on the other, we examined both an unloaded movement cycle, in which the lifting motion was imitated and a loaded movement cycle, in which the box was actually grasped and lifted. Investigation of the kinematics of both cycles provided insight into how the horizontal body CoM position was regulated with respect to the base of support during the forward and backward bending of the upper body. Comparison of the kinematic patterns of the loaded v. unloaded cycle demonstrated a strategy to preserve equilibrium when a load mass was added to the body.

During both movement cycles, a kinematic synergy was revealed, that resembled the synergy described for forward and backward trunk bending movements in stance (Crenna et al. 1987; Oddsson 1988; Oddsson and Thorstensson 1986). The lifting task synergy comprised a simultaneous forward displacement of the head and shoulder, backward displacement of the pelvis and forward displacement of the knee during the downward phase and resulted in a minimal forward displacement of the body CoM. We concluded that the inter-joint coordination was aimed at regulating the CoM position with respect to the base of support, but not at maintaining it on a certain location. The only difference in angular motion between the loaded and unloaded movement cycle was observed in the ankle joint: the onset of ankle extension at the start of the upward phase tended to occur earlier in the loaded cycle. This suggests that the extra postural adjustments related to load pick-up emerged primarily in the ankle joint motion and not in the angular changes of the other joints with a postural function, the knee and hip joint. This is intriguing, because it suggests that, although the ankle, knee and hip all served both a focal and a postural role, the angular changes at the ankle joint appeared to have the most pronounced postural function. This function did not clearly emerge, however, due to a large inter-individual variation and/or too small additional demands on equilibrium control. Therefore, we will further assess the role of the angular changes at the ankle joint in the coordination between movement and equilibrium in a bimanual, whole-body lifting task in the present chapter. We will augment the demands on equilibrium control by letting the subjects perform the same lifting task on a base of support that is reduced to 0.092 m in the anterior-posterior direction. In that way we attempt to create a situation in which the hypothesized role of the ankle in the control of associated postural adjustments will emerge more profoundly.

The conclusion that the angular changes at the ankle joint and muscles crossing the ankle joint have an important function in the control of equilibrium was drawn previously. Ankle muscles were found to play a major role in the postural adjustments associated with trunk bending (Crenna et al. 1987; Massion et al. 1993; Oddsson 1989; Oddsson and Thorstensson 1987; Pedotti et al. 1989) and arm movements (Aruin and Latash 1995a; Cordo and Nashner 1982; Crenna and Frigo 1991) and in the postural responses to an unexpected perturbation (Burleigh et al. 1994; Horak and Nashner 1986). However, in these movements the ankle muscles only subserved a postural function, not a focal one. In the bimanual lifting task, on the contrary, the ankle muscles serve both a focal and postural role. In that case, the finding that muscles crossing the ankle joint have a more pronounced role in the control of equilibrium than other, potentially also postural, leg muscles would be rather novel.

Performing a voluntary trunk motion in a situation in which the demands on equilibrium control were changed was investigated by Massion et al. (1993) and Pedotti et al. (1989). The first authors found that the synergy between upper body and lower limbs was maintained in micro-gravity conditions, although equilibrium threats were absent and, thus, associated postural adjustments were not necessary. The initial standing position and the joint trajectories during for- and backward bending differed in micro- versus normal gravity conditions, as did the electromyographic (EMG) patterns of the ankle muscles (Massion et al. 1993). Performance of a fast backward trunk motion while standing on a reduced base of support induced a change in gastrocnemius activity in trained gymnasts, but not in an untrained group (Pedotti et al. 1989). The altered strategy resulted in a better performance in terms of the percentage of trials in which the whole movement was completed without falling. Given the results of these two studies, the question rises to what extent the kinematic synergy will be maintained in case the lifting task is executed on a reduced base of support.

And if the synergy is maintained, what changes in joint motions or muscle activity patterns will occur?

This chapter investigated the role of ankle muscles in the coordination between the goal directed movement and the associated postural adjustments in a bimanual, whole-body lifting task. The main questions were (1) to what extent the kinematic synergy, i.e. forward head and shoulder motion, backward pelvis displacement and forward knee motion, is maintained when the lifting task is performed on a reduced base of support, (2) what changes in joint motions and (3) what changes in muscle activity patterns occur in case the synergy persists.

Methods

The same nine subjects repeated the lifting task described in the previous chapter, while they were standing on a reduced base of support. This paragraph will describe the methods that are additional to those described before. The analysis of the lifting task performed on the normal base of support was extended in this chapter to include the EMG activity of eight leg muscles.

General procedure and experimental protocol

The subjects repeated the bimanual, whole-body lifting task, while they were standing on a beam (width {left-right} 0.360 m, height 0.060 m, depth 0.092 m) that reduced the anterior-posterior base of support length to 0.092 m (*beam*-condition). The back of the beam was positioned in line with the lateral malleolus. Performance of the task in this condition was compared to the performance on a base of support of normal size, which amounted to 0.277 (±0.010) m (i.e. shoelength) (*shoes*-condition). The subjects wore shoes and adopted the same stance width in each condition. They were instructed to perform the task on the beam, as they had done on shoes, including the unloaded down- and upward movements before grasping and lifting the box. The vertical distance between the base of support and the bottom of the box handles was increased with 0.060 m, to correspond with the distance during lifting on shoes. They executed the reduced base of support task also eight times in a row, separated by a 20-s pause. The subjects were not allowed to practice the lifting task standing on the beam, to exclude any adaptation to the altered support conditions before the actual recordings. To prevent injuries in case of a fall during lifting on the beam, a skilled 'catcher' (a physical therapist with experience in catching patients during gait training) stood at the subjects' left side. Furthermore, subjects were instructed to drop the box if necessary to regain equilibrium.

Apparatus and data acquisition

In addition to the registration of the vertical force exerted on the box and of the sagittal plane position of the fourteen markers attached to the body and the box, the EMG activity was telemetrically obtained from eight superficial leg muscles on the left side of the body: the Tibialis Anterior (TA), Soleus (SOL), Gastrocnemius Lateralis (GL), Gastrocnemius Medialis (GM), Semitendinosus (ST), Gluteus Maximus (GLU), Vastus Lateralis (VL) and Rectus Femoris (RF). Pairs of Ag/AgCl surface electrodes (Medi-Trace pellet electrodes ECE 1801, lead-off area 1.0 cm^2, centre to centre electrode distance 2.5 cm) were attached after standard skin preparation (Basmajian 1978). The electrodes were positioned in longitudinal direction above either the bulkiest part or the middle of the muscle belly. For SOL the electrodes were positioned on the medial edge where the muscle protrudes below GM (Gregoire et al. 1984). To reduce movement artifacts in the EMG signal, the electrodes, connecting wires and pre-amplifiers were carefully taped to the subject's skin and further covered by a skin suit. EMG signals were pre-amplified and transmitted to a Biomes-80 receiver (Glonner electronic GmbH). The EMG signal was sampled (600 Hz, 12 bits) and stored by the VICON system in synchrony with the recorded movements and the box force signals. Off-line, the EMG signal was rectified, after removal of the offset and the frequency was reduced from 600 to 60 Hz by averaging every 10 successive samples. Before the beginning of the first lifting task on shoes, subjects performed Standard Isometric Contractions (SIC) for each muscle group while EMG activity was recorded, according to the procedures described by Gregoire et al. (1984) and Jacobs and Ingen Schenau (1992b). For each muscle group, subjects were asked to perform three maximal contractions for 3 s, while external resistance was provided to prevent segmental movements. Rest periods between SICs were 10 s. The SIC level for each muscle group was calculated as the mean of the processed EMG signal during a period of 1 s in which the signal remained constant. EMG signals obtained during the lifting trials were normalized to 100% SIC level.

Data analysis and statistics

To obtain a measure of the projection of the body CoM onto the ground, the horizontal CoM position was expressed as a percentage of the horizontal distance between the markers at the ankle and fifth metatarsophalangeal (mtp5) joint (CoM_{rel}, 0% at ankle). To compare the relative CoM projection, the trunk inclination and its velocity, the joint angles and angular velocities and the EMG patterns between the two experimental conditions, time traces of the loaded movement cycle of eight trials were averaged per subject and condition. Each trial was synchronized in time to the moment (t=0) the subject started to exert an upward force on the box. A total of 1183 ms (600 before and 583 ms after t=0) was taken into analysis, according to the duration of the movement cycle that (of the 144 cycles analysed) was performed most quickly. Next, the parameter curves of the nine subjects were averaged per condition (mean ± 1 standard error of the mean -SEM-). To investigate the covariation in the angular changes, a Principal Components Analysis (PCA) was applied on the time traces of trunk inclination and hip, knee and ankle joint angles of each movement cycle for each trial and each subject (see chapter 3). The significance of differences between average results was checked with a paired samples t-test and between PCA results with a Wilcoxon test for two related samples (application of a Kolmogorov-Smirnov Goodness of Fit Test proved that some output parameters of the PCA were not a sample from the normal distribution). In each test, an alpha level of 0.05 was employed to determine statistical significance.

Results

Performance of the lifting task on a reduced base of support

All subjects were able to execute the lifting task on the reduced base of support without major problems. Small corrective motions to maintain equilibrium were sometimes seen, but large corrections, such as stepping off the beam to prevent falling were never observed. Although subjects had no prior experience with the task, a rather smooth movement execution was already seen in the first trial. No substantial improvement appeared to occur over the next seven trials. Performing the task on the beam took on average a little more time than performing the same task on shoes, although subjects were encouraged to adopt the same speed of movement. The duration of the downward phase was not significantly longer (unloaded cycle: *beam* 904 (±69) ms v. *shoes* 873 (±68) ms, t=2.04, p=0.075 and loaded cycle: *beam* 941 (±98) ms v. *shoes* 904 (±75) ms, t=1.12, p=0.297), but the duration of the upward phase was, though only in the loaded movement cycle (unloaded: *beam* 910 (±82) ms v. *shoes* 884 (±45) ms, t=1.09, p=0.308 and loaded: *beam* 1118 (±138) ms v. *shoes* 1027 (±106) ms, t=2.68, p=0.028).

The kinematic synergy during performance of the lifting task on a reduced base of support
During lifting on the beam, the kinematic synergy resembled the synergy that was observed during lifting on shoes (Figure 4.1). In the downward phase of both conditions, the head, shoulder and Th1 markers shifted forward, the pelvis moved backward and the knee joint marker forward. The opposite pattern was observed during the upward phase in both cases.

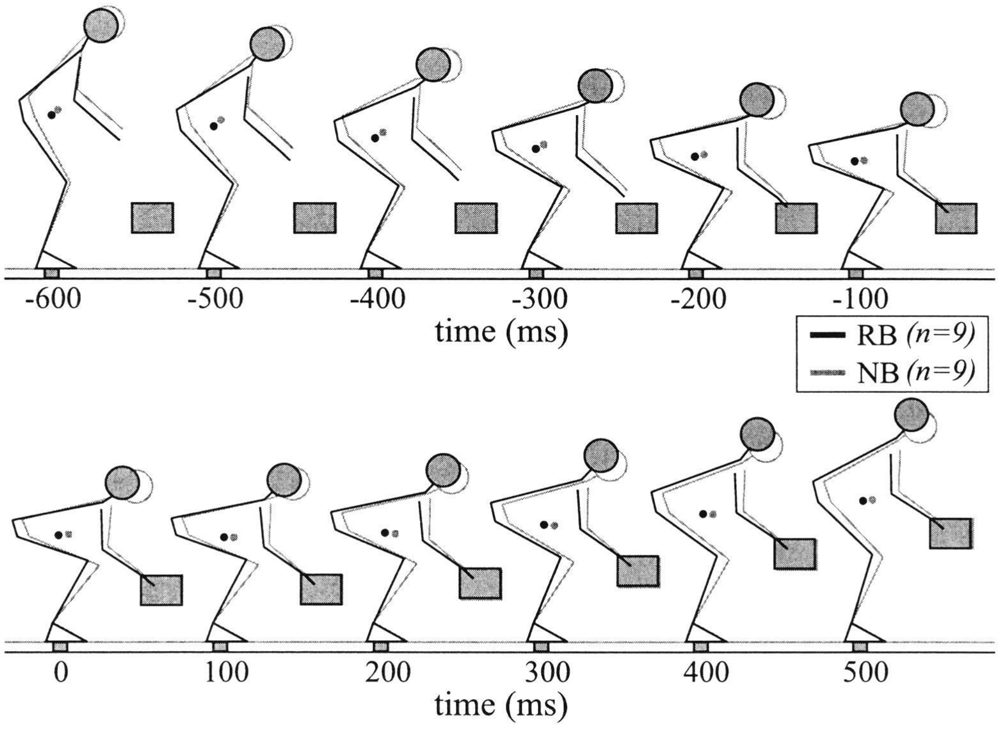


Figure 4.1
Averaged movement of the nine subjects during the lifting task performed on the normal base of support (NB) and on the reduced base of support (RB). The stick-figures are constructed from the average position of the segment ends at each instant in time. The location of the centre of mass (including the load after pick-up) is indicated by a dot. The floor level in the stick-figures of the NB condition was raised 0.06 m, to equal the upper side of the beam in the RB condition. At t=0, subjects started to exert an upward force on the box.

Parameters that give an impression of the control of the horizontal CoM position were similar in both conditions. (1) The maximal horizontal CoM excursion (i.e. the range of motion within one phase) was not significantly different, both during the downward (0.023

(±0.005) m *beam* v. 0.025 (±0.014) m *shoes*, t=0.32, p=0.757) and upward phase (0.037 (±0.004) m *beam* v. 0.046 (±0.013) m *shoes*, t=2.13, p=0.066). (2) The horizontal CoM velocity during a whole movement cycle was not significantly different either: the maximum forward velocity was 0.044 (±0.019) $m \cdot s^{-1}$ on the beam v. 0.065 (±0.029) $m \cdot s^{-1}$ on shoes (t=1.75, p=0.118), the maximum backward velocity was 0.089 (±0.012) $m \cdot s^{-1}$ on the beam v. 0.080 (±0.029) $m \cdot s^{-1}$ on shoes (t=1.06, p=0.321).

Differences between both conditions could be observed, though, in the range of motion of the joint angles and in the horizontal position of the CoM (Figure 4.1). On the beam, the ankle joint angle was larger (i.e. more extension) during both phases and during the upward phase, the knee joint angle seemed slightly larger. No differences appeared to be present in trunk inclination and hip joint angle. Furthermore, the *beam*-CoM was positioned posterior to the *shoes*-CoM throughout the movement. At first sight, the more extended ankle joint angle on the beam could be responsible for the more posterior positioned CoM, because less forward rotation of the whole body around the ankle would indeed yield a more backward positioned CoM. The adaptation of the horizontal CoM trajectory to the reduced base of support was certainly effective, since an unadjusted CoM_{rel} trajectory (that on shoes) would have crossed the front margin of the base of support. Figure 4.2 shows that the *shoes*-CoM_{rel} shifted beyond the front of the beam, while the *beam*-CoM_{rel} fluctuated around the middle of the beam. The front and back of the beam were positioned at, respectively, 84.9 ± 3.2% and 1.1 ± 6.0% of the horizontal ankle-mtp5 distance.

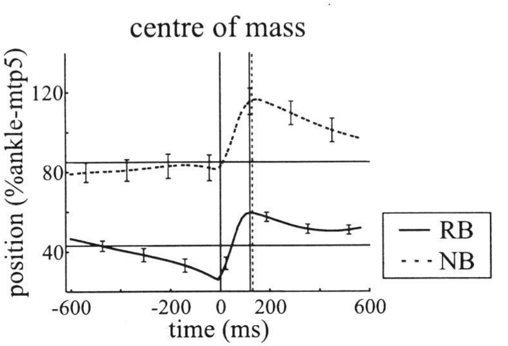


Figure 4.2
Relative position of the centre of mass (expressed as a percentage of the horizontal distance between the markers at the ankle and fifth metatarsophalangeal (mtp5) joint) during the lifting task performed on the normal base of support (NB) and on the reduced base of support (RB). Mean time traces ± 1 SEM (n=9) are presented. The vertical lines mark, from left to right, the onset of the upward force exerted on the box (t=0) and the moment of lift-off of the box in the RB (solid) and NB (dashed) condition. The upper and lower horizontal lines indicate respectively the front (85%) and middle (43%) of the beam in RB.

Main adaptation in joint motions and muscle activity patterns during the lifting task on a reduced base of support

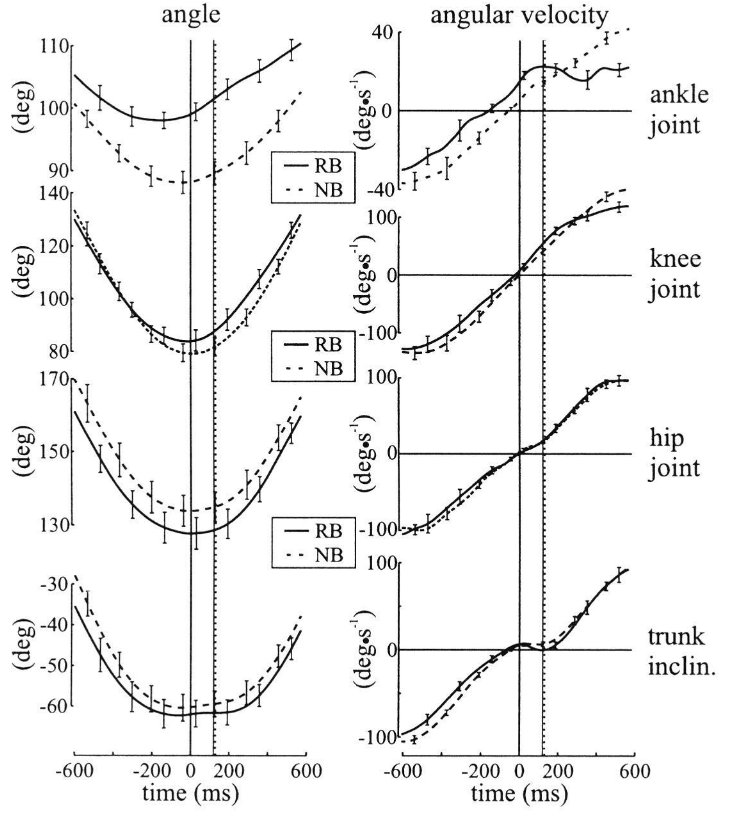


Figure 4.3
Angle and angular velocity in the ankle, knee and hip joints and in the trunk inclination with respect to the vertical, during the lifting task performed on the normal base of support (NB) and on the reduced base of support (RB). Mean time traces ± 1 SEM (n=9) are presented. The vertical lines mark, from left to right, the onset of the upward force exerted on the box (t=0) and the moment of lift-off of the box in the RB (solid) and NB (dashed) condition.

Comparison of both conditions with respect to the time traces of trunk inclination and its velocity and of joint angle and angular velocity at ankle, knee and hip, shows that the alteration in base of support conditions affected the ankle joint most (Figure 4.3). This joint showed less flexion throughout the movement and the angular velocity was smaller (in absolute values) during a large part of both phases on the beam. Note that the ankle extension started about 100 ms earlier in the *beam*-condition, long before the onset of the upward phase. The (absolute) smaller angular velocity for the knee and hip joints and trunk inclination is in agreement with the observed longer duration of task performance when lifting on the beam. Small differences in knee and hip joint angle and in trunk inclination were observed between both conditions. For the hip joint angle and trunk inclination, these differences were already present at the onset of the downward phase and remained more or less constant during task execution.

In agreement with the changes in ankle angle and angular velocity, considerable adjustments in EMG pattern of muscles crossing the ankle joint were found (Figure 4.4). During lifting on the beam, the TA was more active in the first part of the downward phase, but less active in the latter part, while the Triceps Surae (TS, i.e. SOL, GL and GM) showed more activity throughout this phase. The combined change in activity of the ankle muscles opposed the forward rotation of the lower leg more during lifting on the beam, yielding the smaller flexion velocity and the earlier onset of an extending velocity of the ankle joint shown in Figure 4.3. In the upward phase, TA activity was larger and TS activity smaller on the beam, which resulted in the smaller extending velocity of the ankle joint from 200 ms onwards (Figure 4.3). EMG levels of RF, VL and GLU showed some small differences between both conditions, but large inter-individual variations within each condition suggest that these differences would not reach a level of significance. The ST activity was considerably higher during the latter part of the upward phase on the beam, a finding that can be related to the lower extending velocity of the knee joint towards the end of the upward phase (Figure 4.3).

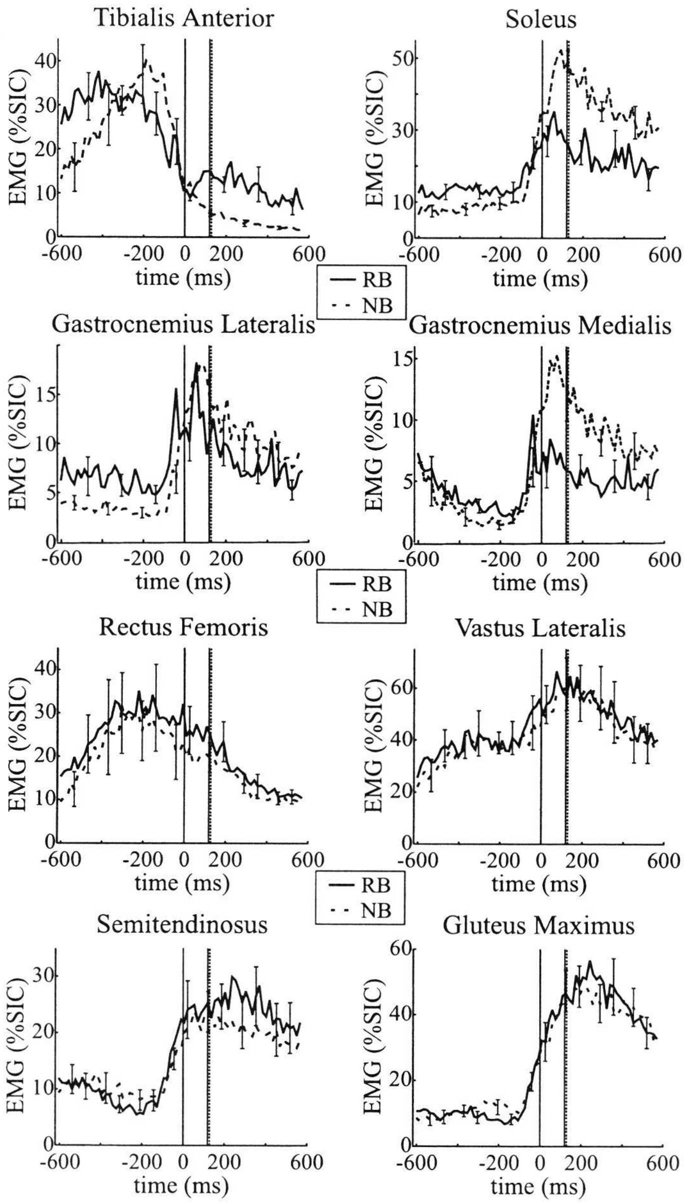


Figure 4.4
EMG patterns of eight leg muscles, during the lifting task performed on the normal base of support (NB) and on the reduced base of support (RB). Mean time traces ± 1 SEM (n=9) are presented. The vertical lines mark, from left to right, the onset of the upward force exerted on the box (t=0) and the moment of lift-off of the box in the RB (solid) and NB (dashed) condition.

The general alteration in EMG pattern of the ankle muscles during lifting on the beam could already be identified in the first *beam*-trials, although not equally well in all subjects (Figure 4.5). Subject 6 was the only one who did not clearly display the general change in pattern shown in Figure 4.4. However, his *beam*-EMG patterns were different from his *shoes*-EMG patterns. Subject 5 (representing five subjects) demonstrated large parts of the general change in TA and TS activation: increased SOL and GL activity during the downward phase and increased TA, combined with decreased TS activity during the upward phase. Subject 8 (representing three subjects) clearly showed the general adaptation in EMG pattern of TA and TS: decreased TA and increased TS activity during the downward and increased TA, combined with decreased TS activity during the upward phase.

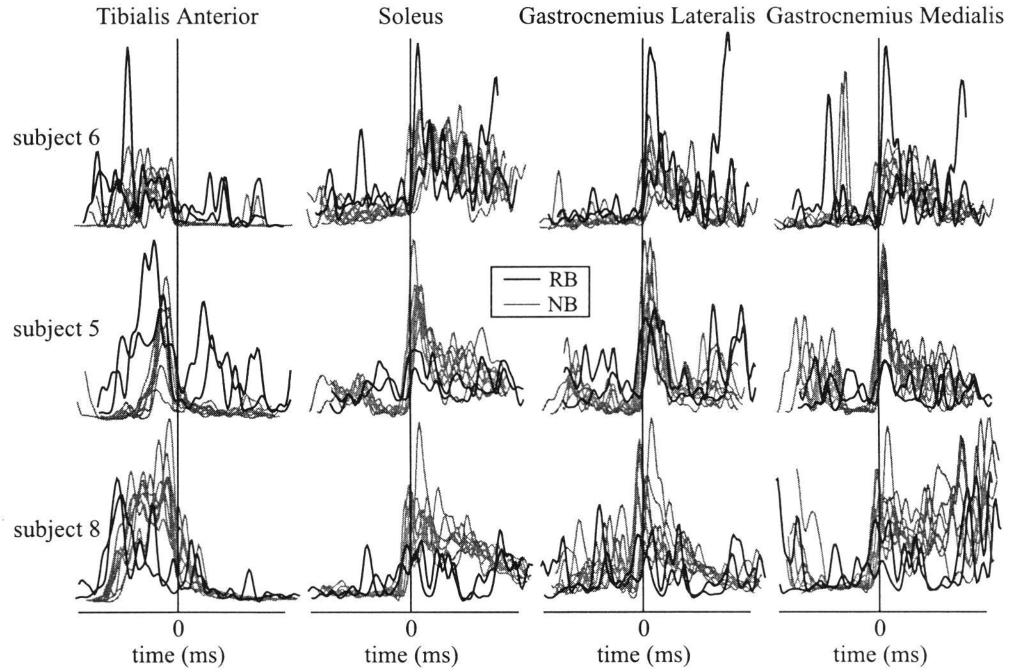


Figure 4.5
EMG patterns of four muscles crossing the ankle joint, during the loaded movement cycle of all eight trials on the normal base of support (NB, grey time traces) and of the first two trials on the reduced base of support (RB, black time traces) for three subjects. The time traces were filtered (unlike the time traces of Figure 4.4) with a digital low-pass filter (6 Hz, 2nd order Butterworth, zero phase lag). The vertical line marks the onset of the upward force exerted on the box (t=0). All time traces were synchronized to this instant. The duration of the complete loaded movement cycle was for all three subjects about 2 s, ± 0.9 s before and ± 1.1 s after t=0.

During lifting on the beam, the time traces of the knee and hip joint angles kept the smooth U-shape that was observed during lifting on shoes (Figure 4.6). Furthermore, the invariance in time traces of the knee and hip joint angles over two movement cycles and eight consecutive trials within one subject was present in both the reduced and normal base of support condition. The observed difference between the loaded and unloaded *beam*-curves is due to the longer duration of the loaded v. unloaded upward phase (this subject: unloaded 948 (±51) ms v. loaded 1102 (±69) ms, t=5.59, p=0.001). The *beam*-time traces of the trunk inclination and, in particular, the ankle joint angle did no longer show the smooth U-shaped pattern, especially not in the loaded cycle. Furthermore, both parameters displayed a little more variation over consecutive trials, compared with the parameters during lifting on shoes. Moreover, one *beam*-trial of this particular subject did not follow the general pattern at all. In this trial, the fourth, a small disturbance of equilibrium was compensated through an adjustment of the ankle joint angle (the deviating trunk inclination curve is the expression of the equilibrium disturbance), while the knee and hip joint angles were not adjusted. The earlier onset of ankle extension in the loaded v. unloaded cycle that was revealed during lifting on shoes was even more pronounced during lifting on the beam (all subjects: -71 (±59) ms in the loaded v. +36 (±56) ms in the unloaded cycle, t=8.96, p<0.001). Furthermore, the onset of ankle extension in the loaded upward phase occurred earlier on the beam than on shoes, though not significantly earlier (all subjects: *beam* -71 (±59) ms v. *shoes* -21 (±55) ms, t=2.07, p=0.072).

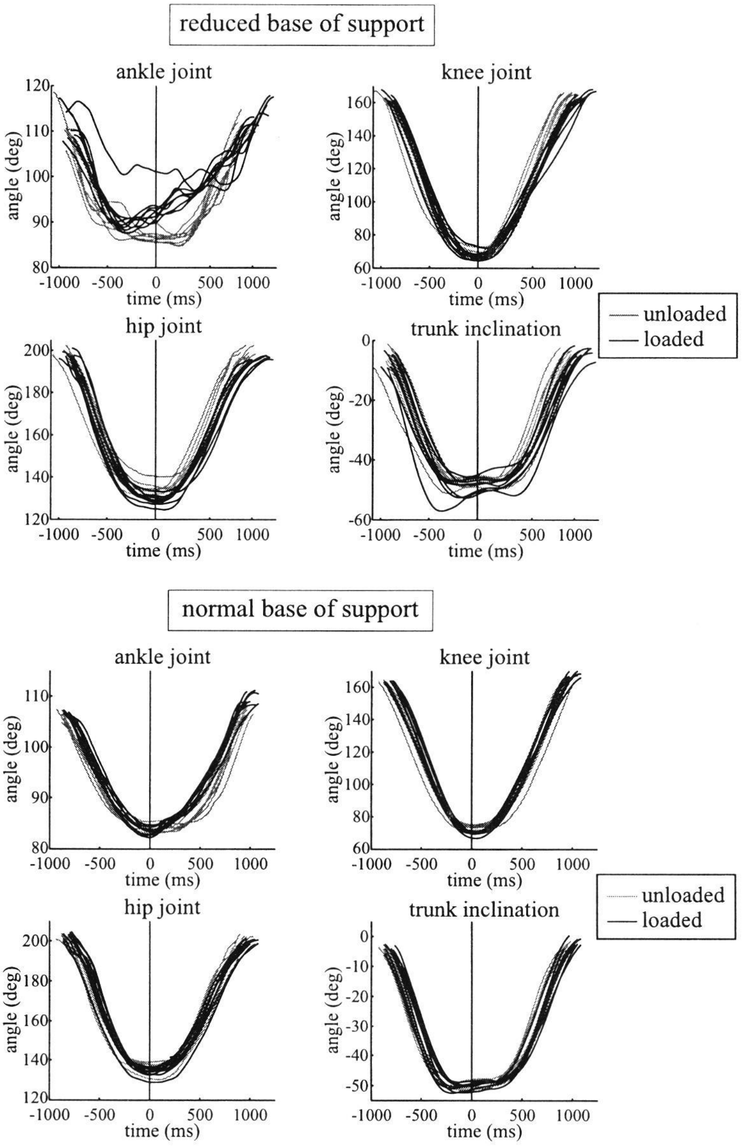


Figure 4.6
Time traces of the angles in ankle, knee and hip joints and of the trunk inclination relative to the vertical during the unloaded and loaded movement cycle of eight consecutive trials of one subject, for the reduced base of support condition (upper panels) and the normal base of support condition (lower panels). Each time trace is presented from the onset of the downward to the end of the upward phase. The curves are synchronized in time to the onset of the upward force on the box (t=0) and the corresponding moment in time in the unloaded movement cycle.

Table 4.1
Results of the Principal Components Analysis (PCA), applied on the time traces of the trunk inclination and the hip, knee and ankle joint angles during the unloaded and loaded movement cycle of the lifting task performed on a reduced base of support (RB). The PCA was performed for each subject and each trial separately, yielding seventy-two cases per movement cycle. Median values and range of values (minimum-maximum) are presented for the percentage of variance in the ensemble of the four angles that could be accounted for by each of the four Principal Components (PC1 to PC4) and for the percentage of variance in the individual angles that could be accounted for by the first Principal Component (PC1). To enable comparison, median values of the eight parameters are included in parentheses for the lifting task performed on a normal base of support (NB).

	unloaded movement cycle			loaded movement cycle		
	median	range RB	(median NB)	median	range RB	(median NB)
% explained variance by:						
PC1	98.25	74.70 - 99.60	(99.10)	93.75	77.00 - 99.00	(98.20)
PC2	1.45	0.30 - 25.20	(0.65)	5.65	0.80 - 22.30	(1.50)
PC3	0.15	0.00 - 0.70	(0.10)	0.40	0.10 - 1.80	(0.20)
PC4	0.00	0.00 - 0.10	(0.00)	0.00	0.00 - 0.10	(0.00)
% explained variance by PC1 in:						
ankle	98.12	3.17 - 99.84	(99.40)	92.55	54.06 - 99.49	(98.30)
knee	99.83	99.19 - 99.98	(99.90)	99.59	97.73 - 99.89	(99.80)
hip	99.61	97.50 - 99.97	(99.80)	98.99	97.03 - 99.83	(99.60)
trunk inclin.	99.03	88.67 - 99.92	(99.40)	96.93	82.42 - 99.78	(98.70)

The results of the PCA (Table 4.1), applied on the individual time traces of trunk inclination and hip, knee and ankle joint angles of all subjects confirmed the observations regarding the particular subject presented in Figure 4.6. In both movement cycles, the covariation of the angular changes was lower on the beam than on shoes: the median percentage of variance in the ensemble of the four angles explained by the first Principal Component (PC1) was 98.25% v. 99.10% in the unloaded cycle (Z=-5.609, p<0.001) and 93.75% v. 98.20% in the loaded one (Z=-6.976, p<0.001). Furthermore, the covariation of the angular changes on the beam proved to be lower in the loaded, compared with the unloaded movement cycle: the median percentage of variance in the ensemble of the four angles attributed to PC1 was respectively 93.75% and 98.25% (Z=-6.209, p<0.001). In all cases, the reduction in percentage of variance in the ensemble of the four angles explained by PC1 could be ascribed to a reduction in the percentage of variance in the ankle joint angle that was explained by PC1 (Table 4.1, compare unloaded and loaded cycle, NB and RB). Thus, the covariation between the angular changes at the ankle joint on the one hand and the changes at the knee and hip joint and trunk inclination on the other was reduced in case equilibrium constraints were increased, that is when picking up a load and when performing the task on a reduced base of support.

Discussion

Maintenance of the kinematic synergy during the lifting task on a reduced base of support

The synergy that was revealed during performance of a bimanual, whole-body lifting task on an unrestricted base of support (chapter 3), was in essence maintained when the task was executed on a base of support that was reduced in the anterior-posterior direction. In the downward phase of both conditions, the forward motion of the head and shoulders and the downward movement of the whole body (the goal directed action) was accompanied with a backward pelvis displacement (the associated postural adjustment). The opposite pattern was observed during the upward phase in both conditions. Thus, maintenance of the global character of the kinematic synergy during trunk movements was found when the demands on equilibrium control were increased (this study) and when those demands were reduced (micro-gravity conditions, Massion et al. 1993). However, in both cases relatively minor, but relevant adjustments in the trajectory or range of motion of one or more joints occurred.

In the present study the global character of the kinematic synergy was maintained on a short base of support. On the contrary, Horak and Nashner (1986) demonstrated that the postural strategy in response to an unexpected perturbation to equilibrium of standing subjects was considerably adjusted to changed support surface conditions. The "ankle" strategy (rotating the whole body primarily around the ankle joints) usually displayed on a normal base of support was replaced by a pure "hip" strategy (bending the upper body at the hip joints) or a mixed "ankle-hip" strategy on a short base. This change in strategy was necessary, because the high ankle torques, that are required to return the passively displaced CoM over the base of support, can not be exerted on a short base of support. A high extending or flexing ankle torque can only be exerted if the centre of foot pressure (CoP) can be positioned close to, respectively, the toes or heels (Okada and Fujiwara 1984) and, per definition, the CoP can not be positioned at locations that don't contact the base of support. The experimental conditions in the study of Horak and Nashner (1986) differed from those in our study: the equilibrium perturbation was passively induced v. self-inflicted and the time available to counteract the perturbation was less than 500 ms v. several seconds. In the first case, the environmental and task constraints were such that the system had no degrees of freedom left within the "ankle" strategy to react to the sudden equilibrium perturbation, while in the latter case, apparently, enough degrees of freedom were left within the observed synergy and sufficient time was available to prospectively counteract the perturbation.

A key role for the ankle joint and ankle muscles in the control of movement and equilibrium

A considerable adjustment in range of motion of the ankle joint yielded a very effective adaptation of the trajectory of the CoM with respect to the reduced base of support throughout the movement. Less flexion in the ankle, compared with the *shoes*-condition (Figure 4.3), yielded a more backward positioned L5-S1 joint marker and, in that way, the CoM projection was kept close to the ankles, fluctuating around the middle of the beam. Only minor adjustments were observed at the other joints. Furthermore, at the start of the loaded upward phase, the onset of ankle extension occurred earlier on the beam than on shoes (Figure 4.6). Thus, the increased challenge to equilibrium maintenance imposed by performing the lifting task on a reduced base of support on the one hand and picking up a load in that situation on the other, was mainly encountered by adjustments in range of motion of the ankle joint. Moreover, a sudden perturbation to equilibrium during lifting was apparently also counteracted by an adjustment in the ankle joint motion only, as shown in the kinematics of the deviating trial in Figure 4.6. The adjustments in range of ankle joint motion suggest that information about the horizontal CoM position relative to the new base of support was available at a central level to adjust the motor commands. Forssberg and Hirschfeld (1994) showed that the support surface size was taken into account in the composition of the postural response of sitting adults in reaction to platform perturbations. The responses were small when the perturbation moved the upper part of the body over the large base of support in front of the hips, but large in case the upper body was moved in the other direction. They suggested that the postural control system sets the threshold of the postural responses according to an internal representation of the body, that includes the relation between CoM and support surface (Forssberg and Hirschfeld 1994).

Changes in EMG pattern of muscles crossing the ankle joint accomplished the change in range of ankle joint motion on the beam, thus providing support for our hypothesis that ankle muscles played a more important role in the control of associated postural adjustments in this lifting task than the muscles crossing the other leg joints with a postural function. Only minor changes in activity of muscles crossing the knee and hip joints were observed. Adaptation of ankle muscle activity patterns to altered equilibrium constraints was also found during fast trunk movements performed under micro-gravity conditions (Massion et al. 1993) and in fast backward trunk bending while standing on a short (0.05 m) base of support (Pedotti et al. 1989). Furthermore, in other than trunk movements, the associated postural adjustments in the legs were reduced in magnitude or completely absent when additional support was provided to the body, that is when the challenge to equilibrium maintenance was reduced (Cordo and Nashner 1982; Friedly et al. 1984; Nardone and Schieppati 1988). Massion (1992) proposed two mechanisms, that could underlie the changes in muscle synergy when alterations in postural support conditions have to be encountered: (1) a short term learning process which changes the previous synergy and creates a new one and (2) an immediate adaptation of the muscle synergy, based on the interaction between centrally organized muscle activity patterns and sensory cues that inform the subject about the altered conditions. The second mechanism was proposed to govern the modification of the postural adjustments for step initiation that was found when stepping was accompanied with an externally imposed change in body position (Burleigh et al. 1994). In the present study, the changes in EMG pattern of the ankle muscles were observed as soon as the *beam*-condition was encountered (Figure 4.5), suggesting that short term learning can be ruled out and that the process of adaptation was based on sensory feed-back informing the subject about the reduced base of support. The sensory cues could update the internal representation of the interaction between the mechanical properties of the musculo-skeletal system and environment, such that the predictive motor plan is altered (Hirschfeld and Forssberg 1991; 1992). Although the *beam*-condition was novel to all subjects, they had about 5 s to experience the new length of the base of support in the first trial, because they performed several unloaded down- and upward movements before the actual recording began. They were unexperienced, though, in lifting the load on the beam.

Ankle joint muscles were found to play a major role in the control of equilibrium in various motor acts, e.g. in for- and backward trunk bending (Crenna et al. 1987; Massion et al. 1993; Oddsson 1989; Oddsson and Thorstensson 1987; Pedotti et al. 1989), in the initiation of forward oriented movements, like stepping (Burleigh et al. 1994; Crenna and Frigo 1991), in the postural responses to an unexpected perturbation to equilibrium in stance (Burleigh et al. 1994; Horak and Nashner 1986) and in the anticipatory postural adjustments before picking up a load (chapter 5). The question rises as to what mechanism underlies the important function of the ankle muscles in the control of equilibrium? Application of a biomechanical analysis of the lifting task provides insight into the effects of a change in torque at the ankle joint on the dynamics of a (moving or stationary) body. A change in ankle torque directly alters the position of the CoP (Okada and Fujiwara 1984). Thus, from a global perspective, regarding the whole body as a single free body, the point of application of the ground reaction force vector can be translated, without necessarily affecting the magnitude and direction of that vector, that is, without affecting the current horizontal and vertical acceleration of the body CoM. However, a shift in position of the CoP relative to the position of the body's CoM, does alter the external moment (i.e. the moment applied by the ground reaction force on the CoM), inducing a change of the angular momentum of the whole body (Toussaint et al. 1995). From a local perspective, regarding the lower leg segment as a single free body, a change in ankle torque alters the velocity of the for- or backward rotation of the lower leg (provided that the knee torque remains unchanged) and, thus, the velocity of the for- or backward sway of the whole body around the ankle joint. In global terms, this coincides with a change of the whole-body angular momentum (Toussaint et al. 1995). An active displacement of the CoP was found in the anticipatory postural adjustments prior to picking up a load (chapter 5; Toussaint et al. in press 1997a; in press 1997b) and in the postural responses to an unexpected perturbation to equilibrium during the performance of a lifting task (Toussaint et al. submitted). This mechanism presumably also underlay the initiation of forward oriented movements (Burleigh et al. 1994; Crenna and Frigo 1991; Nardone and Schieppati 1988) and of for- and backward trunk bending (Crenna et al. 1987; Pedotti et al. 1989). In view of the present results and the reviewed literature, we suggest that the active displacement of the CoP position is important in the control of posture and equilibrium during goal directed actions. Given the mechanical effect of activity of muscles crossing the ankle joint on CoP position, these muscles indeed may play a crucial role in the control of associated postural adjustments.

Acknowledgements
The authors would like to thank Idsart Kingma, Michiel de Looze, Jaap van Dieën, Gerrit Jan van Ingen Schenau and Helga Hirschfeld for reviewing the manuscript.

Chapter 5

Anticipatory postural adjustments in a bimanual, whole-body lifting task with an object of known weight

Abstract

Anticipatory postural adjustments (APA) were studied in a dynamic multi-joint movement, in which the legs serve both a focal and a postural role. Eight male subjects bimanually lifted a barbell (20% of body mass) after several unloaded movement cycles using two distinct lifting techniques. Picking up a load induces a perturbation to balance, because the centre of mass (CoM) of the combined body and load shifts forward at the moment of load pick-up. Furthermore, the inertia of the load decelerates the backward rotation of the body towards the erect posture. Both perturbations were found to be counteracted by APA in kinematics, kinetics and leg muscle activity patterns. The APA were characterized by an increase in the backward directed horizontal CoM momentum, an increase in the backward directed whole-body angular momentum and a forward shift of the centre of foot pressure (CoP) for both techniques. Anticipatory adjustments in activity of muscles crossing the ankle joint were shown to control the CoP position, which was important to accomplish the required combination of anticipatory changes in horizontal and angular momenta. The APA in kinematics and kinetics were modulated according to the dynamic requirements of each lifting technique, although it could have been expected that picking up and lifting the same load at equal speed would have yielded a similar perturbation in each condition.

Introduction

The performance of voluntary multi-joint movements is often accompanied by a displacement of the body's centre of mass (CoM) with respect to the base of support. Since this implies disequilibrium (Gahery and Massion 1981), the central nervous system (CNS) has to control the movements of the multi-segmented body in such a way that the task goal is reached without a major disturbance of equilibrium (Crenna et al. 1987). Several studies have demonstrated that the expected perturbation to balance elicits *anticipatory postural adjustments* (APA) (Bouisset and Zattara 1987), defined as actively initiated movements which predictively counteract disturbances of balance that are associated with a voluntary movement (Belen'kii et al. 1967; Cordo and Nashner 1982; Lee et al. 1987). In forward trunk bending, for example, the forward shift of the upper-body CoM was compensated by a backward displacement of the lower-segments CoM, prior to the perturbing trunk movement (Oddsson 1990). The APA proved to be specific for the direction of the forthcoming voluntary movement (Bouisset and Zattara 1987; Nashner and Forssberg 1986; Oddsson 1990). Furthermore, the anticipatory kinematic, kinetic and muscle activity patterns were found to be modulated according to the dynamic requirements of the motor task in which the voluntary movement and subsequent perturbation were induced (Hirschfeld and Forssberg 1991; Nashner and Forssberg 1986).

In a bimanual, whole-body lifting task equilibrium maintenance is challenged by the pick-up of an object in front of the body. The addition of an extra mass to the body causes the CoM to shift forward with respect to the base of support and the inertia of the grasped object will tend to decelerate the backward rotation of the whole body towards the final erect posture. The latter event does not disturb equilibrium in the sense that the CoM is displaced with respect to the base of support, but it could impede a smooth extending motion of the whole body if not adequately anticipated. APA can be expected to counteract the adverse effects of the forward CoM shift and deceleration of the extension movement prior to their occurrence, analogous to the APA that were revealed prior to an expected perturbation, like raising the arm(s) (Belen'kii et al. 1967; Bouisset and Zattara 1987). If APA are present in this lifting task, they should occur prior to load pick-up, during the downward movement of the lifter towards the object. Thus, the anticipatory changes in leg muscle activity have to be integrated with the muscle activity patterns that generate (or decelerate) the downward movement. This implies that "focal" (Cordo and Nashner 1982) leg muscles will have a "postural" (Cordo and Nashner 1982) role too. Until now, studies on APA have mainly investigated tasks that were characterized by a clear distinction between focal and postural segments or joints, e.g. forward and backward trunk bending (Crenna et al. 1987; Oddsson 1988; Oddsson and Thorstensson 1986; Pedotti et al. 1989) and arm movements during upright stance (Friedly et al. 1984; Lee et al. 1987; Riach et al. 1992; Zattara and Bouisset 1988). A few studies investigated tasks in which the legs first served a postural and then a focal function, e.g. rising on tip-toes or flexing one leg (Mouchnino et al. 1992; Nardone and Schieppati 1988; Rogers and Pai 1990) and only a handful of studies examined tasks in which the focal segments had at the same time a postural function, e.g. pulling or pushing a handle during locomotion (Hirschfeld and Forssberg 1991; 1992; Nashner and Forssberg 1986; Patla 1986). A bimanual, whole-body lifting task allows examination of the integration of APA in an ongoing multi-joint movement, in which the legs serve both a focal and a postural function.

APA were indeed revealed in bimanual load lifting and they seemed to be aimed at minimizing the destabilizing effect of load pick-up (Commissaris and Toussaint 1995). The main anticipatory changes in kinematics involved an increase in the backward directed horizontal CoM momentum and an increase in the backward directed whole-body angular momentum prior to load pick-up. These anticipatory changes should be reflected in the ground reaction force ($\boldsymbol{F}_g$), because the rate of change of the horizontal momentum equals the horizontal component of $\boldsymbol{F}_g$ and the rate of change of the angular momentum equals the moment effect of $\boldsymbol{F}_g$ about the CoM (defined as the external moment, Toussaint et al. 1995). Control of the magnitude and direction of $\boldsymbol{F}_g$ was, next to control of joint angular changes, considered crucial in the execution of *contact force control tasks*, tasks that are characterized by the exertion of a force on the environment (Doorenbosch and Ingen Schenau 1995; Ingen Schenau et al. 1992; Jacobs and Ingen Schenau 1992a). Since the control of the contact force relies on the generation of a particular *combination* of net torques at the joints involved (Ingen Schenau et al. 1992), it was suggested that the CNS does not control the *individual* joint torques, but rather the contact force (Jacobs and Macpherson 1996; Macpherson 1988a; 1988b). The execution of the APA in bimanual load lifting can be conceived as a contact force control task, in that a force of certain magnitude and direction must be generated by the feet against the ground to accomplish the anticipatory changes in horizontal and angular momenta. Thus, it is suggested that $\boldsymbol{F}_g$ is one of the key variables under control of the CNS during the APA prior to load pick-up. Besides magnitude and direction, the point of application of $\boldsymbol{F}_g$ (the centre of foot pressure, CoP) has to be controlled, because a significant role was ascribed to the CoP in accomplishing the required combination of horizontal and angular momenta in a bimanual lifting task (Toussaint et al. 1995).

The present study investigated the anticipatory $\boldsymbol{F}_g$ pattern (magnitude, direction and CoP) and the relation between the global $\boldsymbol{F}_g$ pattern and local muscle actions during the APA. More generally, we studied the strategies employed by the CNS to counteract balance perturbations brought about by picking up a load in bimanual lifting. An expected perturbation to balance was induced by picking up a rather heavy load (20% of body mass), positioned just within reach in front of the subjects. Two lifting techniques were applied, that differed with respect to the forward-backward body CoM displacement. In the leglift (straight back, bent legs) this displacement was much smaller than in the backlift (bent back, straight legs) (Commissaris and Toussaint 1995). Thus, a modulation of the anticipatory kinematic, kinetic and muscle activity patterns, according to the dynamic requirements of each lifting technique was hypothesized.

Methods

Subjects

Eight healthy male subjects (mean age 22.3 ± (standard deviation) 1.5 years, body height 1.79 ± 0.07 m, body mass 71 ± 11.7 kg, footlength 0.267 ± 0.067 m) participated in the experiments, after they had given written informed consent and after approval of the Faculty's ethical committee. None of the subjects reported a history of low-back disorders or other motor impairments.

Experimental procedures

After several unloaded down- and upward movements, the subjects were asked to pick up the barbell, to lift it in a straight vertical line and to come to a full stop holding the barbell at acromion height (Figure 5.1). The barbell was placed in front of the toes, at such a distance (heel-barbell 0.615 ± 0.054 m) that the subject was just able to pick it up. The desired vertical trajectory of the barbell was indicated by two flexible metal wands that were positioned in front of the barbell (from the subject's perspective) at the left and right end. In the lowest position, the vertical distance between the load and the ground was standardized at 14% of body height. Care was taken to further standardize the execution of the lifting movements to minimize inter-subject variability. The subjects were asked (1) to restrict their movements to the sagittal plane, (2) to keep the heels on the ground at load pick-up and (3) to guard their balance throughout the movement. Balance was considered to be disturbed when the heels lost contact with the ground or when a compensatory step was made to prevent falling.

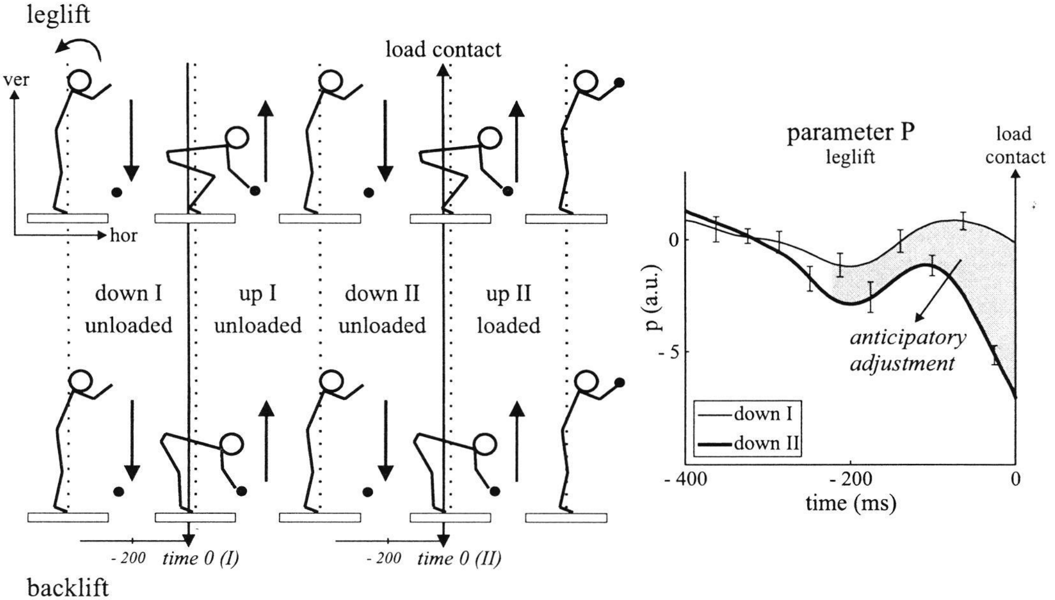


Figure 5.1
Schematic outline of the bimanual lifting task (left) and definition of anticipatory adjustment (right). After several unloaded down- and upward movements, the subjects picked up the barbell, lifted it in a straight vertical line and came to a full stop holding the barbell at acromion height (left panel). Two lifting techniques were examined: the leglift (upper sequence) and backlift (lower sequence). The last two movement cycles (unloaded down I - unloaded up I - unloaded down II - loaded up II) were recorded. The dotted vertical lines mark the end of a downward (lowest vertical body centre of mass position) or upward phase (highest vertical body centre of mass position). The right solid vertical line marks the moment of load contact in down II (t=0) and the left one marks the corresponding instant in time in down I. Kinematics, kinetics and muscle activity patterns of down II (after which *the load* was picked up) and down I (after which *no load* was picked up) were compared from 400 ms before t=0 to t=0. A significant difference in, for example, parameter P (right panel) between down II (thick line) and I (thin line) is defined as an *anticipatory adjustment* (filled area).

In each technique condition (leglift, backlift), the subjects performed six successive trials in a range of lifting speeds; the duration of the last downward phase ranged from 1.0 to 0.4 s. The movement speed was imposed with the help of auditory cues generated by a metronome. To attune to the desired speed, subjects performed several movement cycles without picking up the load. When the required rhythm was attained, one of the authors counted down to the moment of barbell lift-off, starting at the beginning of the penultimate downward phase (left stick-figure in Figure 5.1). The order of the adopted lifting technique was randomized. Each set of trials took about 15 min and subjects were allowed a pause of at least 15 min between the two sets. All subjects performed three to five practice trials in each technique condition to familiarize themselves with the experimental task.

Anticipatory postural adjustments

To identify the anticipatory movement, force and muscle activity patterns within the patterns normally found in a downward motion, the kinematics, kinetics and EMGs of a downward phase (I, Figure 5.1) after which *no load* was picked up were compared with the same parameters of the next downward phase (II, Figure 5.1) after which *the load* was picked up. Thus, both downward phases were *unloaded*, but differed from each other with respect to the expected perturbation to balance that would occur after downward phase II, but not after I. A difference in parameter curves between both phases would then be indicative of APA, directed at counteracting the expected perturbation. Figure 5.1 shows an anticipatory adjustment for parameter P; before load contact, the parameter values in downward phase II (thick line) decrease relative to the values of downward phase I (thin line). The difference (filled when significant) is defined as APA. Note that the moment of load contact defines the end of the time period in which parameter curves of down II and I are compared. This moment always precedes the end of downward phase II, i.e. the lowest vertical position of the body CoM.

Anthropometry

Reflective markers (ø 25 mm) were attached to the subject's right body side to indicate the location of the fifth metatarsophalangeal joint, the ankle joint (distal part of the lateral malleolus), the knee joint (lateral epicondyle), the hip joint (greater trochanter), the lumbo-sacral (L5-S1) joint (as in Looze et al. 1992), the spinous process of the first thoracic vertebra (Th1), the head (caput mandibula), the lateral border of the acromion, the elbow joint (lateral epicondyle), the wrist joint (ulnar styloid) and the hand (a small stick attached to the third metacarpus). An additional marker was attached to the right end (from the subject's perspective) of the barbell. The coordinates of the acromion marker were used to determine the position of the shoulder joint. The length of the base of support was inferred from markers placed on the heel and the distal end of the most prominent toe. The coordinates of the joint positions defined eight body segments in the sagittal plane: the feet, lower legs, upper legs, pelvis, trunk/head, upper arms, forearms and hands(/load). Anthropometric data (body mass, length of segments) were measured. The mass of each segment and the positions of segmental CoMs, except for the trunk, and moments of inertia were calculated according to Plagenhoef et al. (1983) and Looze et al. (1992). The mass and location of CoM of the hands were adapted at the instant the hands grasped the load to include the mass and location of the CoM of the load. The coordinates of the markers on Th1 and the L5-S1 joint were used to determine the position of the trunk CoM during the movement according to an optimization procedure, which improved the estimated trajectory of the body CoM (Kingma et al. 1995).

Kinematics and kinetics

During the last two lowering/lifting cycles (depicted in Figure 5.1) the marker positions were recorded at a frame rate of 60 Hz, using a 3-D semi-automatic video-based motion registration system (VICON™, Oxford Metrics Ltd., four camera set-up). The raw sagittal plane coordinates of these markers were low-pass filtered with a digital filter (zero phase lag, 5 Hz, 2nd order Butterworth). The angle of each segment was calculated relative to the horizontal. Numerical differentiation (Lanczos 5-point differentiation filter) of the time histories of the segment angles and CoM positions yielded (angular) velocities and accelerations. The caudo-cranial and dorso-ventral translational directions and the backward (counter-clockwise) rotational direction were defined positive (Figure 5.1, left panel). Vertical and anterior-posterior components of $\boldsymbol{F}_g$ were recorded by means of a strain-gauge force platform (1.0 by 1.0 m). The analog force signals were amplified, low-pass filtered (30 Hz, 4th order), sampled (60 Hz, 12 bits) and stored in synchrony with the movement registration by the VICON system. From the distribution of the force components, the CoP of the force vector was calculated (with a maximal error of 3 mm). Both components of $\boldsymbol{F}_g$ and the CoP were low-pass filtered with a digital filter (zero phase lag, 5 Hz, 2nd order Butterworth).

Electromyography

The electromyogram (EMG) of eight superficial leg muscles on the left side of the body were telemetrically obtained from the following muscles: Tibialis Anterior (TA), Soleus (SOL), Gastrocnemius Lateralis (GL), Gastrocnemius Medialis (GM), Semitendinosus (ST), Gluteus Maximus (GLU), Vastus Lateralis (VL) and Rectus Femoris (RF). Pairs of Ag/AgCl surface electrodes (Medi-Trace pellet electrodes ECE 1801, lead-off area 1.0 cm^2, centre to centre electrode distance 2.5 cm) were attached after standard skin preparation (Basmajian 1978). The electrodes were positioned in longitudinal direction above either the bulkiest part or the middle of the muscle belly. For SOL the electrodes were positioned on the medial edge where the muscle protrudes below GM, according to Gregoire et al. (1984). To reduce movement artifacts in the EMG signal, the electrodes, connecting wires and pre-amplifiers were carefully taped to the subject's skin and further covered by a skin suit. EMG signals were pre-amplified and transmitted to a Biomes-80 receiver (Glonner electronic GmbH). The EMG signal was high-pass filtered at 7 Hz to reduce the amplitude of possible movement artifacts, full-wave rectified, smoothed using an analog 20 Hz 3rd order low-pass filter and further amplified on-line by special purpose equipment. The signal was then sampled (60 Hz, 12 bits) and stored in synchrony with the movement registration by the VICON system. Before the beginning of the first trial, subjects performed Standard Isometric Contractions (SIC) for each muscle group while EMG activity was recorded, according to the procedures described by Gregoire et al. (1984) and Jacobs and Ingen Schenau (1992b). For each muscle group, subjects were asked to perform three maximal contractions for 3 s, while external resistance was provided to prevent segmental movements. Rest periods between SICs were 10 s. The SIC level for each muscle group was determined by taking the mean of the processed EMG signal for a period of 1 s in which the signal remained constant. The EMG signals obtained during the lifting trials were normalized to 100% SIC level.

Biomechanical analysis

The nature of the APA and the relation between the global $\boldsymbol{F}_g$ pattern and local EMG patterns during the APA was assessed using a mechanical analysis of the movement. This analysis relies on the interdependence of local and global mechanics. Local muscle contractions will change the segmental linear and angular momenta that, summed over all segments, equal the change in whole-body linear and angular momenta. On a global level, the magnitude and direction of $\boldsymbol{F}_g$ are the net result of local muscle contractions which generate torques at multiple joints (Ingen Schenau et al. 1992; Toussaint et al. 1992).

Considering the whole body as a single free body, the external $\boldsymbol{F}_g$ (given gravity) changes the linear momenta of the body CoM and the external moment changes the body's angular momentum (Toussaint et al. 1995). Thus, control of the magnitude and direction of $\boldsymbol{F}_g$, through an adequate combination of joint torques, is assumed important to accomplish the anticipatory changes in linear and angular momenta. Furthermore, control of the CoP position, through activity of muscles crossing the ankle (Crenna and Frigo 1991; McIlroy and Maki 1993; Okada and Fujiwara 1984), is assumed crucial in accomplishing the required combination of anticipatory changes in linear and angular momenta.

The following global biomechanical parameters were calculated:

- The instantaneous horizontal and vertical momentum of the CoM of the whole body, including the load after pick-up, calculated from the sum of respectively the horizontal and vertical momenta of all segmental CoMs (i.e. calculated from kinematic and anthropometric data). The change in horizontal momentum should equal the horizontal component of $\boldsymbol{F}_g$ (force platform data). The change in vertical momentum should equal the vertical component of $\boldsymbol{F}_g$ (force platform data) minus the subject's weight.
- The instantaneous angular momentum of the whole body, including the load after pick-up, calculated from the sum of the segmental angular momenta according to Toussaint et al. (1995) (i.e. calculated from kinematic and anthropometric data). The change in angular momentum should equal the external moment, generated by $\mathbf{a} \times \boldsymbol{F}_g$ (force platform data), with **a** the vector from CoM to CoP.
- The horizontal positions of CoM and CoP were expressed as a percentage of the base of support length, i.e. the horizontal heel-toe distance (with 0% at the heel).
- The magnitude and direction of $\boldsymbol{F}_g$, with the direction calculated from the angle of $\boldsymbol{F}_g$ with respect to the ground (0° forward horizontal, 90° upward vertical).

At a local level, the net sagittal plane joint torques at ankle, knee and hip were calculated using an inverse dynamics approach (Looze et al. 1992). The calculation started at the feet using force platform data as input. Torques having a plantar flexing or extending effect at the joint were given positive values.

Data analysis and statistics

To identify APA prior to the expected perturbation (i.e. load pick-up), the biomechanical parameters and EMGs of downward phase II were compared with those of down I (Figure 5.1). The moment of load contact (t=0) defined the end of the time period of comparison.

The duration of this period was 400 ms, determined by the trials with the shortest duration of the downward phases (533 ms). Load contact was set at the instant at which the velocity with which the hand marker approached the barbell marker reached zero. The moment of lift-off of the barbell was defined at the instant at which the vertical displacement of the barbell exceeded 2.5 mm. To define a 't=0' in down I, the time difference between hand-barbell contact and the lowest vertical body CoM position in down II was subtracted from the instant of the lowest vertical body CoM position in down I (Figure 5.1).

The 96 trials were marked as 'balance' (62) or 'imbalance' (34). The latter category was excluded from the analysis. To test the significance of differences between parameter curves of down II and I and to test whether lifting technique influenced the anticipatory kinematics, kinetics and muscle activity patterns, uni- and multivariate analyses of covariance (ANCOVA and MANCOVA respectively) with repeated measures were applied, with anticipation (down I, down II) and lifting technique (leglift, backlift) as within-subject factors and with the duration of the downward phase in each of four conditions as a covariate. From the 62 balance trials, sixteen pairs of leglift and backlift trials (two pairs for each subject) were selected for analysis. The leg- and backlift trials were matched according to the duration of downward phase II, with a maximum difference in duration of 50 ms. All parameters within each condition were checked for normality, using a Kolmogorov-Smirnov test. Significant overall effects were further examined with ANCOVAS and paired samples t-tests were applied to determine the onset of significant differences between parameter curves of down II and I for each technique separately. Effects were considered to be significant at $p<0.05$.

Results

The global anticipatory postural adjustments

To examine whether APA were present and, if so, influenced by lifting technique, an ensemble of eight global biomechanical parameters was compared between downward phase II and downward phase I. One sample (17 ms) prior to the moment of load contact defined the baseline for this comparison. The hypothesis of normal distribution was not rejected for any of the parameters. The MANCOVA indeed revealed a significant effect for anticipation, while the interaction between technique and anticipation was significant too (Table 5.1). The covariate was not significantly related to any of the dependent variables. The univariate F-

tests with respect to the main effect 'anticipation' and the interaction effect 'anticipation x technique' are presented in Table 5.2. It should be noted that of the three features of $\boldsymbol{F}_g$ (magnitude, direction and CoP) that were assumed important in the control of APA, the magnitude did not show an anticipatory adjustment.

Table 5.1
Repeated measures multi-variate analysis of covariance on eight global biomechanical parameters, with the duration of the downward phase in each condition as a covariate.

Effect	Wilk's lambda	F-value	p-value
anticipation	0.056	14.783	0.001 *
within cells regression (covariate)	0.501	0.870	0.580
lifting technique	0.013	67.490	0.000 *
within cells regression (covariate)	0.300	2.037	0.182
anticipation x technique	0.069	11.859	0.002 *
within cells regression (covariate)	0.596	0.592	0.761

* Effects significant at $p<0.05$.

Table 5.2
Univariate F-tests for the effects 'anticipation' and 'anticipation x technique' on angular momentum ($\boldsymbol{L}$), position of CoM and CoP relative to the base of support (BoS) (CoM_{rel} and CoP_{rel}), external moment ($\boldsymbol{M}_{ext}$), horizontal and vertical linear momentum ($\boldsymbol{p}_{hor}$ and $\boldsymbol{p}_{ver}$) and magnitude and direction of $\boldsymbol{F}_g$ ($\boldsymbol{F}_{mag}$ and φ).

Dependent variable		Main effect: Anticipation				Interaction effect: Anticipation x Technique			
		MS effect	MS error	$F(1,14)$	p-value effect	MS error	MS	$F(1,14)$	p-value
$\boldsymbol{L}$	($kg{\bullet}m^2{\bullet}s^{-1}$)	34.36	2.97	11.58	0.004 *	0.01	1.68	0.01	0.928
CoM_{rel}	(% BoS)	0.72	24.61	0.03	0.867	109.94	3.62	30.36	0.000 *
CoP_{rel}	(% BoS)	987.27	22.16	44.55	0.000 *	199.10	12.41	16.04	0.001 *
$\boldsymbol{M}_{ext}$	($N{\bullet}m$)	463.18	91.89	5.04	0.041 *	932.92	72.17	12.93	0.003 *
$\boldsymbol{p}_{hor}$	($kg{\bullet}m{\bullet}s^{-1}$)	18.97	6.87	2.76	0.119	27.29	4.87	5.60	0.033 *
$\boldsymbol{p}_{ver}$	($kg{\bullet}m{\bullet}s^{-1}$)	1.47	60.92	0.02	0.879	19.86	18.80	1.06	0.321
$\boldsymbol{F}_{mag}$	(N)	2747.14	3197.71	0.86	0.370	3088.22	3103.43	0.99	0.335
φ	(deg)	28.32	0.77	36.65	0.000 *	4.64	0.91	5.11	0.040 *

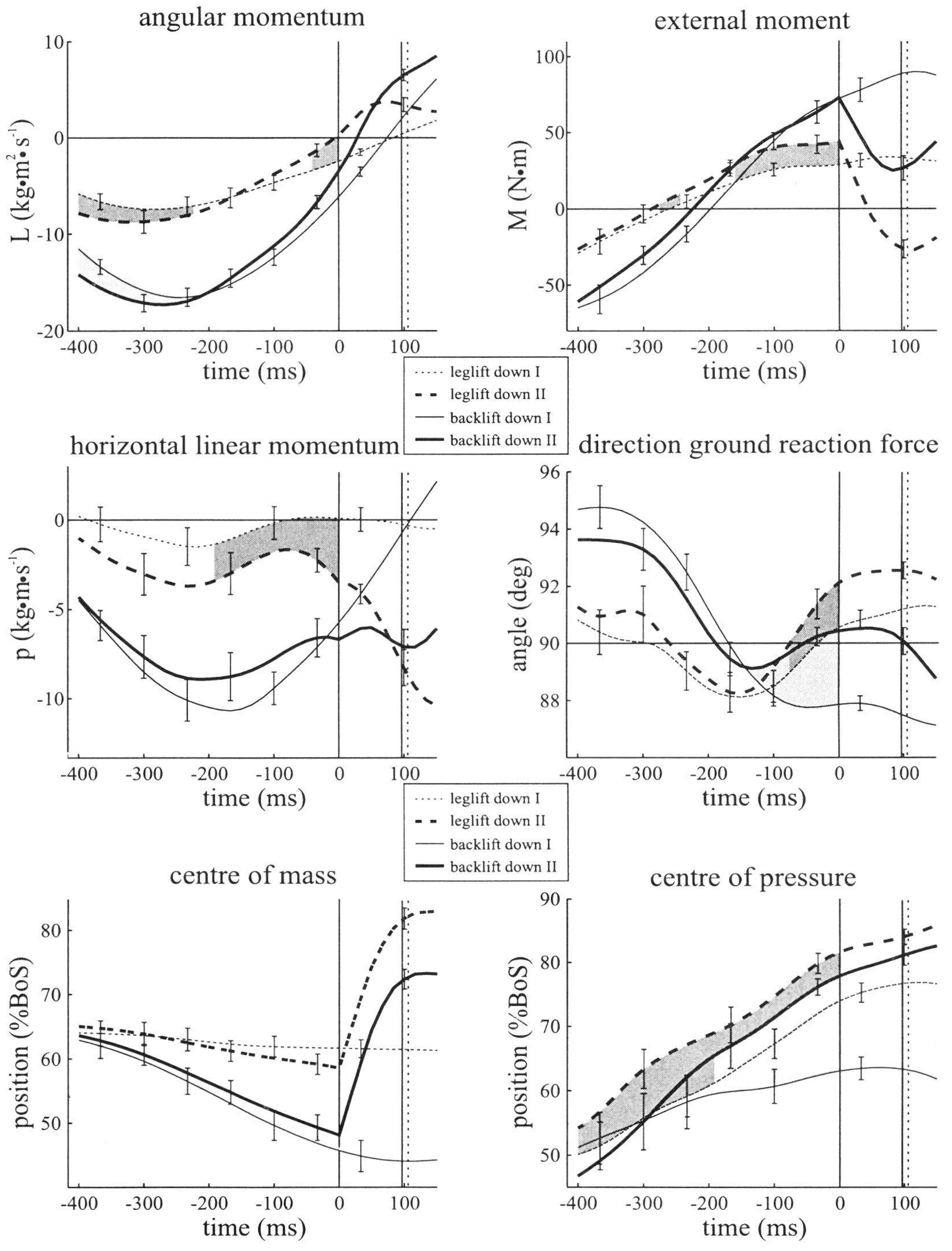
* Effects significant at $p<0.05$.

Next, the factor time was included in the analyses, to identify the onset of a significant difference between parameter curves of down II and I for each technique separately. Only parameters with significant 'anticipation' and/or 'anticipation x technique' effects were further analysed. Per parameter, an ANOVA was performed with factors anticipation (down I, down II) and time (sample 1 to 24, i.e. time -400 to -17 ms), followed by a paired samples *t*-test. The angular momentum showed an anticipatory increase (i.e. less forward rotational velocity) in the last 50 ms of down II (Figure 5.2, upper left panel) for both techniques. This suggests that the expected deceleration of the backward body rotation after load pick-up was indeed anticipated. As the change in angular momentum equals the external moment, anticipatory changes in angular momentum should be accompanied by anticipatory changes in the external moment in both technique conditions. The early increase in external moment in down II (leglift -275 to -242 ms and backlift -308 to -175 ms, upper right panel) was thus related to the decreasing difference in angular momentum between down II and I (starting around -300 ms for both techniques) and to the end of the first anticipatory change in angular momentum around -220 ms. The late anticipatory increase in external moment for the leglift was related to the anticipatory increase in angular momentum close to the moment of load contact.

The horizontal momentum displayed a pronounced anticipatory adjustment for the leglift, but no significant change in backlifting (Figure 5.2, middle left panel). In the first case, the backward directed momentum was increased (from -190 ms onwards), most likely to counteract the expected forward CoM shift at load pick-up. For the backlift, a similar increase in backward directed momentum occurred, though only after load contact. Since differences between down II and I in the period from load contact to lift-off can be attributed to both anticipatory adjustments and the mechanical effect of load pick-up, we can not conclude with certainty that the increase in backward directed momentum during backlifting was an APA. As the change in horizontal momentum equals the horizontal component of $\boldsymbol{F}_g$, the anticipatory change in leglift horizontal momentum should be accompanied by an anticipatory change in the direction of $\boldsymbol{F}_g$. Indeed, a significant increase the direction of $\boldsymbol{F}_g$ (i.e. $\boldsymbol{F}_g$ vector more backward directed in down II) was observed during leglifting in the last 75 ms before load contact (middle right panel, dark gray area). For the backlift, however, the $\boldsymbol{F}_g$ vector changed direction too in down II, from -125 ms onwards (light gray area). This anticipatory increase was related to a decreasing difference in horizontal momentum between down II and I, however, a significant anticipatory change in horizontal momentum did not occur before the moment of load contact.

The relative CoM position did not significantly change in down II relative to down I, both for leg- and backlifting (Figure 5.2, lower left panel). Apparently, the expected perturbing *forward* CoM shift was not counteracted by an anticipatory *backward* CoM displacement, but by an anticipatory increase in backward velocity (for the leglift). Note the quick forward CoM shift, about 25% of the footlength (67 mm), between load contact and lift-off in both technique conditions. Both during leg- and backlifting, the CoP was positioned considerably more towards the toes in down II than in I (lower right panel). For the leglift trials, this difference was already present 400 ms before load contact, suggesting that the adjustment had started shortly after the beginning of the downward phase.

Figure 5.2
Time traces of the downward and part of the upward phase of a bimanual lifting task for the angular momentum (upper left), external moment (upper right), horizontal linear momentum (middle left), direction of the ground reaction force (middle right) and position of centre of mass and centre of pressure relative to the base of support (BoS) (lower left and right, respectively) for the leglift (dashed lines) and backlift (solid lines). Mean time trace ± 1 standard error of the mean (SEM) ($n=16$) of pairs of leg- and backlift trials (matched with respect to duration of down II) are shown. The solid vertical line at t=0 ms indicates load contact in down II and the corresponding instant in time in down I. The vertical lines around t=100 ms mark lift-off of the barbell during leglifting (dashed line) and backlifting (solid line). The filled areas (dark gray during leglifting and light gray during backlifting) indicate significant differences between curves of down II and I, i.e. periods of *anticipatory postural adjustments.*


angular momentum
L (kg•m²•s⁻¹)
time (ms)
external moment
M (N•m)
time (ms)
leglift down I
leglift down II
backlift down I
backlift down II
horizontal linear momentum
p (kg•m•s⁻¹)
time (ms)
direction ground reaction force
angle (deg)
time (ms)
leglift down I
leglift down II
backlift down I
backlift down II
centre of mass
position (%BoS)
time (ms)
centre of pressure
position (%BoS)
time (ms)

The local anticipatory postural adjustments

To examine how the APA were reflected in the local parameters, muscle activity patterns and joint torques of downward phase II were compared with those of downward phase I. One sample (17 ms) prior to the moment of load contact defined the baseline for the comparison of joint torques, while a time period of four samples (from -50 ms to 0 ms) was averaged for the comparison of EMGs. Again, the hypothesis of normal distribution was not rejected for any of the parameters. The eleven ANCOVAS revealed a significant effect for anticipation on EMG activity of the ankle plantar flexors SOL, GL and GM and on the torques at the ankle and hip (Table 5.3). Furthermore, the latter parameter was significantly influenced by lifting technique. The covariate was not significantly related to any of the dependent variables.

Table 5.3

Repeated measures univariate analyses of covariance on eleven local parameters, with the duration of the downward phase in each condition as a covariate. Only the effect of 'anticipation' and 'anticipation x technique' on EMG activity of Tibialis Anterior (TA), Soleus (SOL), Gastrocnemius Lateralis and Medialis (GL and GM respectively), Semitendinosus (ST), Gluteus Maximus (GLU), Vastus Lateralis (VL) and Rectus Femoris (RF) and on torques at the ankle, knee and hip ($\boldsymbol{T}_{\text{ankle}}$, $\boldsymbol{T}_{\text{knee}}$ and $\boldsymbol{T}_{\text{hip}}$) are tested.

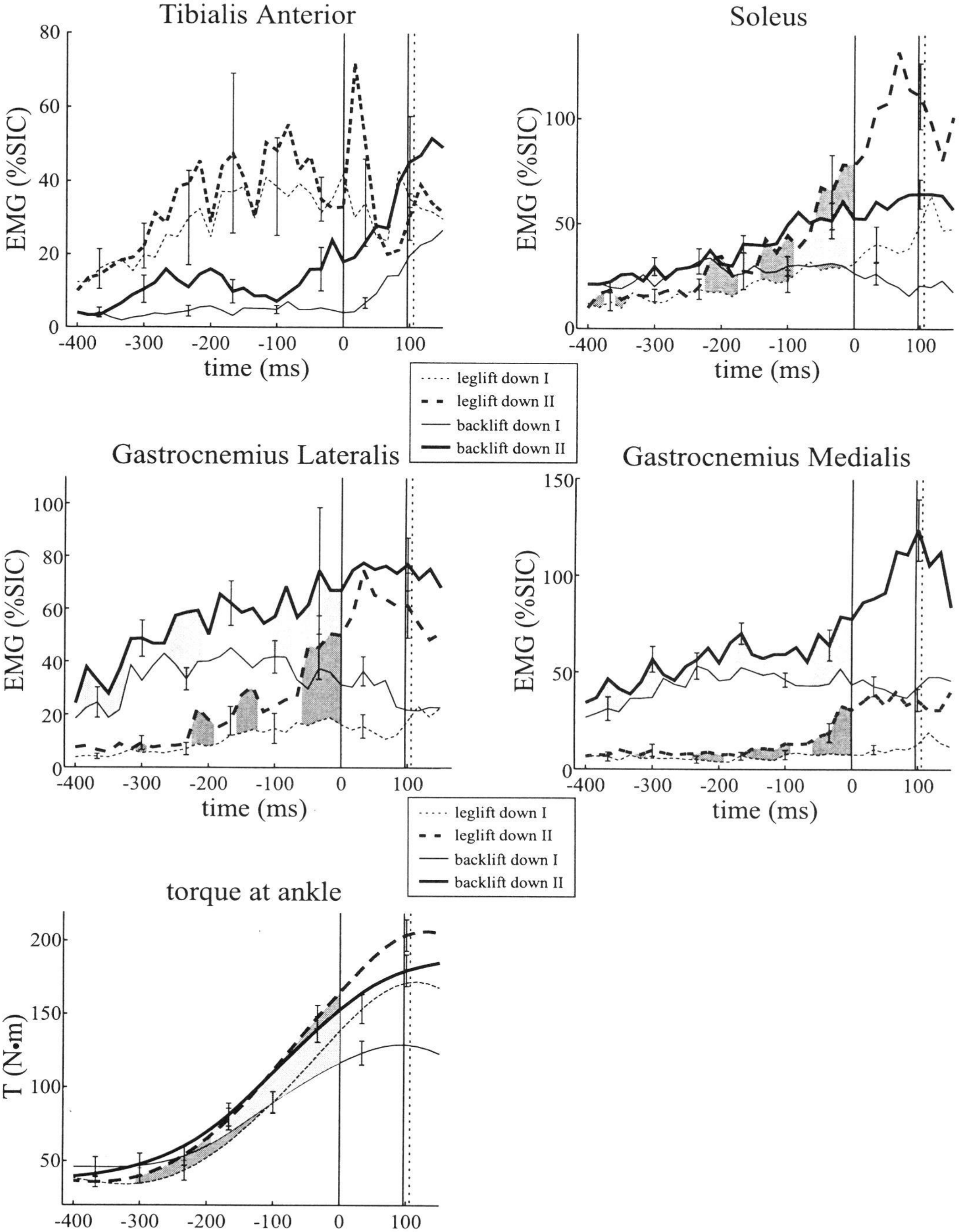
Dependent variable		Main effect: Anticipation				Interaction effect: Anticipation x Technique			
		MS effect	MS error	$F(1,14)$	p-value effect	MS error	MS	$F(1,14)$	p-value
TA	(%SIC)	150.13	258.15	0.58	0.458	1255.80	342.23	3.67	0.076
SOL	(%SIC)	10672.10	805.02	13.26	0.003 *	830.71	272.88	3.04	0.103
GL	(%SIC)	4232.88	778.40	5.44	0.035 *	2.62	801.46	0.00	0.955
GM	(%SIC)	3364.64	269.64	12.48	0.003 *	369.45	333.30	1.11	0.310
ST	(%SIC)	1819.28	613.33	2.97	0.107	386.78	370.24	1.05	0.324
GLU	(%SIC)	10.32	556.53	0.02	0.894	1600.04	583.84	2.74	0.120
VL	(%SIC)	644.29	143.52	4.49	0.052	7.15	95.05	0.08	0.788
RF	(%SIC)	70.62	26.53	2.66	0.125	67.41	54.16	1.25	0.283
$\boldsymbol{T}_{\text{ankle}}$	(N•m)	6379.29	205.65	31.02	0.000 *	306.58	149.64	2.05	0.174
$\boldsymbol{T}_{\text{knee}}$	(N•m)	355.30	209.09	1.70	0.213	506.38	162.40	3.12	0.099
$\boldsymbol{T}_{\text{hip}}$	(N•m)	3634.25	589.70	6.16	0.026 *	3683.83	322.55	11.42	0.004 *

* Effects significant at $p<0.05$.

Next, the factor time was included in the analyses, to identify the onset of a significant difference between parameter curves of down II and I for each technique separately. Parameters with significant 'anticipation' and/or 'anticipation x technique' effects were further analysed, including the EMG activity of VL for which the effect of anticipation almost reached significance (p=0.052). Per parameter, an ANOVA was performed with factors anticipation (down I, down II) and time (-400 to -17 ms), followed by a paired samples t-test. For both techniques, the ankle plantar flexors SOL, GL and GM increased their EMG activity considerably prior to load contact (Figure 5.3). Furthermore, short periods of increased activity were found throughout down II for all three muscles. The nature of the EMG signal, in combination with surface recordings of muscle activity in periods in which the muscles were mainly lengthening, probably caused the 'gaps' in periods of a continuous anticipatory increase in activity. The anticipatory increase in activity of ankle plantar flexors caused an anticipatory increase in the plantar flexing ankle torque for both techniques (Figure 5.3, from -308 and -192 ms onwards for leglift and backlift respectively). This, in turn, corresponds to the anticipatory forward CoP shift (Figure 5.2), since the ankle torque equals the moment effect of $\boldsymbol{F}_g$ with respect to the ankle joint in a situation in which the foot does not move (Toussaint et al. 1992). The moment effect of $\boldsymbol{F}_g$ increased, because the momentarm (the vector from CoP to ankle joint) increased significantly in both technique conditions, while the magnitude of $\boldsymbol{F}_g$ remained at the same level.

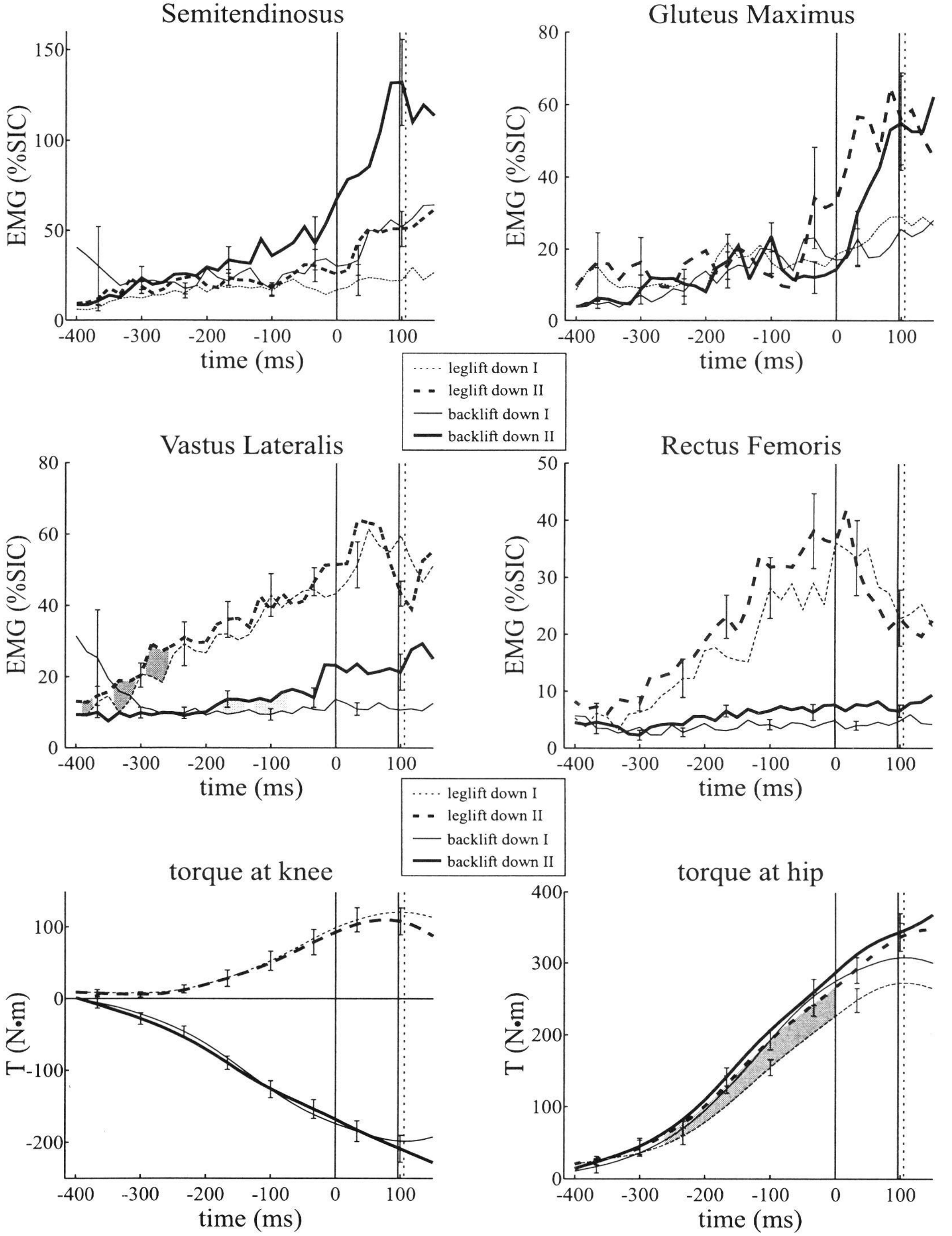
To quantify the relation between the anticipatory activity of ankle muscles (i.e. the *difference* in EMG between down II and I) and the anticipatory shift in CoP position (i.e. the *difference* in CoP between down II and I), a step-wise regression analysis was applied. Independent variables were the anticipatory TA and TS activity (TS: Triceps Surae = SOL+GL+GM) and the dependent variable was the anticipatory CoP shift. The momentarms of TA and TS with respect to the ankle joint rotation centre (Spoor et al. 1990) were accounted for in the effect of muscle activity on ankle torque and on CoP position. An electromechanical delay of 100 ms between the change in EMG and the change in mechanical output (CoP position) was taken into account (Ingen Schenau et al. 1995a). The averaged anticipatory CoP shift ($n=16$) could be fairly well related to the averaged anticipatory activity of TA and TS ($n=16$): the explained variance (R^2) was 62.4% and 88.5% during leg- and backlifting respectively. In leglifting, the explained variance in the anticipatory CoP shift could be completely attributed to the anticipatory TS activity (62.4%), since there was no effect of anticipation on TA EMG (Table 5.3). During backlifting, anticipatory TA activity made a small, but significant contribution to reduce the unexplained variance in the anticipatory CoP shift (from 15.8% to 11.5%). Testing the significance of differences between curves of down II and I (ANOVA and post-hoc paired samples *t*-test) indeed revealed two short bouts of increased TA activity during down II.

Figure 5.3
Time traces of the downward and part of the upward phase of a bimanual lifting task for the muscles crossing the ankle joint (TA, SOL, GL, GM) and the corresponding ankle torque for the leglift (dashed lines) and backlift (solid lines). Mean time trace ± 1 SEM ($n=16$) of pairs of leg- and backlift trials (matched with respect to duration of down II) are shown. Illustrational conventions as in Figure 5.2.


Tibialis Anterior
EMG (%SIC)
time (ms)
Soleus
EMG (%SIC)
time (ms)
leglift down I
leglift down II
backlift down I
backlift down II
Gastrocnemius Lateralis
EMG (%SIC)
time (ms)
Gastrocnemius Medialis
EMG (%SIC)
time (ms)
leglift down I
leglift down II
backlift down I
backlift down II
torque at ankle
T (N•m)
time (ms)

All thigh muscles recorded displayed some increase in activity during down II relative to I, but only the knee extending VL showed a significant anticipatory increase in EMG activity (Figure 5.4). This increase was presumably antagonistic to the anticipatory increase in (knee flexing) activity of the biarticular GL and GM, which was required to displace the CoP forward prior to load pick-up. As a result, the net knee joint torque did not change significantly prior to load contact (Figure 5.4). As no significant anticipatory changes in EMG activity of muscles crossing the hip joint (ST, GLU, RF) was found, the increase in hip extending torque during leglifting (from -258 ms onwards, Figure 5.4) is surprising. Activity of additional muscles, for example, Semitendinosus and Biceps Femoris caput longum may have increased the hip extending torque.

Figure 5.4
Time traces of the downward and part of the upward phase of a bimanual lifting task for the muscles crossing the knee and/or hip joint (ST, GLU, VL, RF) and the corresponding knee and hip torque for the leglift (dashed lines) and backlift (solid lines). Mean time trace ± 1 SEM ($n=16$) of pairs of leg- and backlift trials (matched with respect to duration of down II) are shown. Illustrational conventions as in Figure 5.2.


Semitendinosus
Gluteus Maximus
Vastus Lateralis
Rectus Femoris
torque at knee
torque at hip
EMG (%SIC)
T (N•m)
time (ms)
leglift down I
leglift down II
backlift down I
backlift down II

Discussion

The control of balance in a dynamic multi-joint task

Fascinated by the question how the CNS organizes the coordination of posture and movement, many studies have investigated the control of balance in various motor tasks (e.g. (Cordo and Nashner 1982; Hirschfeld and Forssberg 1992; Kolb and Fischer 1994). For a static situation, like sitting or upright standing, it is commonly accepted that the $\boldsymbol{F}_g$ vector should be continuously aimed at the CoM (Murray et al. 1967; Nashner and McCollum 1985). For a dynamic lifting and lowering task, though, Toussaint et al. (1995) found that during the course of action, the $\boldsymbol{F}_g$ vector pointed substantially in front of or behind the CoM. Hence, $\boldsymbol{F}_g$ exerted a moment about the CoM, defined as the external moment. In line with Winter's view (1995) on the relevance of studying "total body kinetics to learn more about the synergies of human movement", the lifter and load were conceived as one free body diagram and a mechanical whole-body analysis was applied. In that way, it was demonstrated that the presence of an external moment did not disturb balance, but that it was required to accomplish the segmental rotations necessary to reach the task goal, the external moment being equal to the rate of change of the whole-body angular momentum (Toussaint et al. 1995). Thus, in dynamic situations (i.e. in tasks in which the largest part of the body, including the legs, is involved in the primary movement), control of $\boldsymbol{F}_g$ is such that it points away from the CoM whenever necessary (to establish a change in angular momentum), whereas in static situations, control of $\boldsymbol{F}_g$ is such that pointing away from the CoM is minimized. This distinction has to be taken into consideration when the execution of APA is examined in static and dynamic situations. Moreover, it has to be kept in mind that in dynamic multi-joint tasks the leg muscles most likely serve both a focal and a postural role. This can be contrasted with static tasks in which the primary, focal movement is performed by the upper body and in which the leg muscles only have a postural function.

Anticipatory postural adjustments during bimanual load lifting; global patterns

The APA in the bimanual lifting task investigated in the present study were characterized by an increase in the angular momentum (i.e. less forward rotational velocity) and a pronounced increase in the backward directed horizontal momentum (during leglifting only) prior to load contact. These characteristics can be understood from the mechanical considerations that adding a load in front of the body causes the CoM to shift forward and that the inertia of the load decelerates the backward rotation of the body towards an erect posture. It is noteworthy that the perturbing *forward* CoM shift was not counteracted by an anticipatory *backward* CoM shift, but by an anticipatory increase in backward velocity. Presumably the characteristics of the lifting task constrained the possibilities to adjust the horizontal body CoM *position* with respect to the barbell, but not the *velocity*. All this was accompanied by a forward shift of the CoP towards the toes, a crucial event that will be discussed later. These findings suggest that the anticipatory adjustments in kinematics and kinetics of the downward phase before load pick-up were specified in advance on the basis of the expected perturbation. This observation implies a reliance of the feed-forward control on an internal representation of the object's weight, biomechanical properties of the body and specific task elements (Ghez et al. 1991; Hirschfeld and Forssberg 1991; Lacquaniti et al. 1992).

Modulation of the anticipatory postural adjustments according to the lifting technique applied

The global APA were found to be influenced by lifting technique, although it could have been expected that picking up and lifting the same load at equal speed (i.e. a similar perturbation) would have resulted in the same pattern of anticipatory kinematics and kinetics during leg- and backlifting. The differences in APA between both techniques mainly involved differences in onset of the APA (e.g. for the CoP, Figure 5.2) and/or in amplitude of the APA (e.g. for the direction of $\boldsymbol{F}_g$, Figure 5.2). The most pronounced difference was found in the horizontal CoM momentum, which showed an anticipatory increase for the leglift, but not for the backlift (Figure 5.2). It may be concluded that the subjects modulated the global translational and rotational anticipatory adjustments according to the dynamic requirements of each lifting technique, analogous to the step-cycle-phase dependent modulations of anticipatory patterns found during human locomotion (Hirschfeld and Forssberg 1991; 1992; Nashner and Forssberg 1986). However, the modulations in global APA were not reflected in local anticipatory EMG patterns of the muscles recorded: for both technique conditions a distinct anticipatory increase in SOL, GL and GM activity was revealed (Figure 5.3). This may point to a robust part of the anticipatory program or a technique independent synergy (Crenna and Frigo 1991). It can also be suggested that small differences in onset and/or amplitude of the anticipatory muscle activity, responsible for the modulation of the global APA, were not identified by our method of analysis. This suggestion is supported by the finding that the onset of APA in the ankle torque did differ between both techniques (Figure 5.3). Furthermore, other leg muscles than the ones we have recorded EMG activity of could have been responsible for the technique related difference in the global APA.

Anticipatory postural adjustments during bimanual load lifting; relation between global ground reaction force pattern and local muscle activity patterns

On a global level, anticipatory changes in the direction of the $\boldsymbol{F}_g$ vector, in the external moment and in the CoP were found. On a local level, anticipatory changes in activity of leg muscles were demonstrated, accompanied by changes in net joint torques. Ingen Schenau and co-workers have shown that patterns of leg muscle activation in contact force control tasks can not be organized on the basis of the required local joint angular changes alone, since these patterns depend strongly on the magnitude and direction of the global $\boldsymbol{F}_g$ (Ingen Schenau et al. 1992; Jacobs and Ingen Schenau 1992a). Likewise, the patterns of activation of muscles crossing the ankle joint can not be organized on the basis of the required local ankle joint angular changes alone, because the EMG patterns were found to be related to the position of the global CoP (Crenna and Frigo 1991; McIlroy and Maki 1993; Okada and Fujiwara 1984). Ingen Schenau et al. (1995b) argued that detailed internal representations of the properties of the effector system in relation to the environment are indispensable in the control of relatively fast multi-joint contact force control tasks. Thus, we suggest that control of the anticipatory changes in the local parameters requires some representation of the magnitude, direction and point of application of the global $\boldsymbol{F}_g$ vector in relation to the CoM.

Magnitude and direction of the ground reaction force

Although an increase in $\boldsymbol{F}_g$ magnitude prior to load pick-up could have been expected given the downward CoM shift due to the addition of the load mass to the body and given the larger total mass to be accelerated after load pick-up, the APA in global kinematics and kinetics did not involve a significant change in the magnitude of $\boldsymbol{F}_g$. However, a significant anticipatory increase in the direction of $\boldsymbol{F}_g$ was found for both lifting techniques, aimed at counteracting the forward CoM shift at load pick-up. A forward CoM displacement with respect to the base of support implies disequilibrium (Gahery and Massion 1981) and active control of the horizontal CoM position is assumed important in equilibrium regulation (Crenna et al. 1987; Forssberg and Hirschfeld 1994; Horak and Nashner 1986). As the anticipatory muscle activity responsible for the increase in the direction of $\boldsymbol{F}_g$ is concerned, the following strategy may be proposed. The activity of the Triceps Surae increases to generate an extra ankle plantar flexing torque. An increase in activity of the VL neutralizes the side effect of the increase in GL and GM activity on the net knee torque. The other muscles crossing the knee and/or hip joint maintain the same level of activity, such that the net knee and hip torques don't change. This combination of joint torques will, in theory, yield a forward CoP shift, in combination with a more backward directed $\boldsymbol{F}_g$ vector. The proposed combination of anticipatory adjustments in muscle activity and joint torques was indeed demonstrated for backlifting and, for the greater part, also during leglifting (Figures 5.3 and 5.4).

Point of application of the ground reaction force

Previous research has revealed that the CoP serves a crucial role in accomplishing the required combination of horizontal and angular momenta during bimanual lifting tasks (Toussaint et al. 1995). Displacement of the CoP provides a solution to the mechanical 'problem' of changing the horizontal momentum without affecting the angular momentum (and vice versa) (Toussaint et al. 1995). This problem also occurred during the APA in the present study, for a more backward directed $\boldsymbol{F}_g$ vector (i.e. anticipatory increase in direction of $\boldsymbol{F}_g$, Figure 5.2) had to be accomplished without a decrease in the external moment. Figure 5.5 depicts the solution to this problem for the backlift (which was also observed during leglifting). The $\boldsymbol{F}_g$ vector was more backward directed in down II relative to I. Without a forward CoP shift, this would have reduced the momentarm between force vector and CoM (**aII**), because the horizontal CoM position did not significantly change. Hence, the external moment would have been reduced, since the magnitude of $\boldsymbol{F}_g$ did not significantly increase either. However, the momentarm between force vector and CoM did not decrease due to the anticipatory forward CoP shift. Thus, the external moment did not change, while at the same time the $\boldsymbol{F}_g$ vector was more backward directed. The neural strategy proposed in the section above to generate the backward directed $\boldsymbol{F}_g$ vector during the APA provides a solution to the mechanical problem described. This solution resembles the combined "ankle-hip" strategy that was found in automatic postural adjustments following horizontal support surface perturbations (Horak and Nashner 1986). The "ankle-hip" strategy yielded a combined adjustment in $\boldsymbol{F}_g$ direction and CoP position, while a pure ankle strategy was characterized by a CoP displacement and a pure hip strategy by a change in $\boldsymbol{F}_g$ direction.

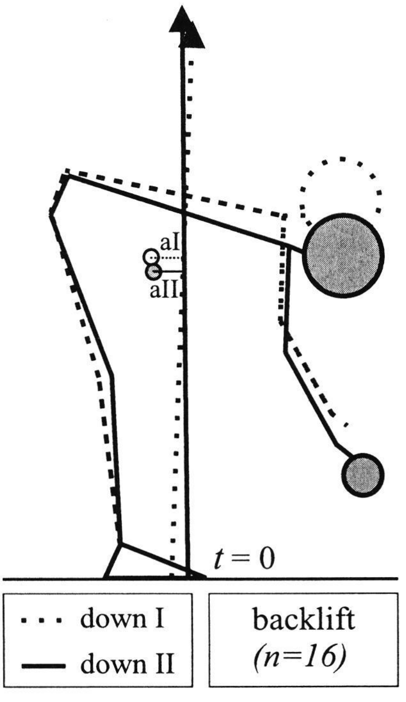


Figure 5.5
Averaged stick-figure of the subject (*16 trials*) at the moment of load contact (t=0) during the task performed with a backlift. Downward phase II (solid lines) is superimposed on the stick-figure of the preceding downward phase I (dotted lines). The ground reaction force vector is shown relative to the location of the centre of mass (dot). The momentarm between vector and centre of mass is indicated by **aI** (for down I) and **aII** (for down II).

The CoP shift was found to be related to activity of ankle muscles. The step-wise regression analyses showed that for the backlift 88.5% of the variance in anticipatory CoP shift could be attributed to the activity of the 4 ankle muscles recorded, while this percentage was 62.4% for the leglift. The remaining part of the variance in CoP shift may be attributed to activity of ankle muscles that were not recorded, like the long toe extensors and flexors. Furthermore, Okada and Fujiwara (1984) observed that the Abductor Hallucis muscle showed an exponential increase in activity as the CoP shifted forward beyond 60% footlength. Since the CoP shifted more beyond 60% footlength during leglifting compared with backlifting (lower right panel, Figure 5.2), absence of the Abductor Hallucis muscle in the regression analysis may have caused the larger percentage of unexplained variance in the anticipatory CoP shift in leglifting compared with backlifting. The finding of a relation between the anticipatory CoP shift and anticipatory ankle muscle activity suggests an active role of ankle muscle activity in control of the CoP position. The significance of the ankle muscles in the control of posture and equilibrium was emphasized before, for example in the postural adjustments associated with trunk bending (Massion et al. 1993; Oddsson 1989; Oddsson and Thorstensson 1987) and arm movements (Aruin and Latash 1995a; Cordo and Nashner 1982;

Crenna and Frigo 1991) and in the postural responses to an unexpected perturbation (Burleigh et al. 1994; Horak and Nashner 1986). However, in these movements the ankle muscles only have a postural function, not a focal one. In the bimanual lifting task, on the contrary, the ankle muscles serve both a focal and postural role.

Acknowledgements

The authors would like to thank Marco Hoozemans and Michiel Ober for their support in data acquisition and processing and for many fruitful discussions. Idsart Kingma, Jaap van Dieën and Gerrit Jan van Ingen Schenau are gratefully acknowledged for reviewing the manuscript.

Chapter 6

Load knowledge affects low-back loading and control of balance in lifting tasks

Abstract

This study investigated the effect of the presence or absence of load knowledge on the low-back loading and the control of balance in lifting tasks. Low-back loading was quantified by the net sagittal plane torque at the lumbo-sacral joint. The control of balance was studied by the position of the centre of mass relative to the base of support, the horizontal and vertical momentum of the centre of mass and the angular momentum of the whole body. In a first experiment, eight male subjects lifted a rather heavy load (22% of body mass), using a leglift and a backlift, while they were familiar with the load mass. To counteract the threat to balance, imposed by picking up a load in front of the body, the subjects performed specific preparations, based upon the known load mass: prior to load pick-up, profound changes in the horizontal and angular momentum were found. The preparations were technique specific. Preserving balance seemed easier while picking up a load with a backlift than with a leglift. In the second experiment, twenty-five male subjects lifted a 6 kg box, which they expected to be 16 kg, because, in a series of lifts, the load mass was changed from 16 to 6 kg without their knowledge. Despite the 10 kg difference in actual load mass, the net torque at the lumbo-sacral joint was not different between lifting 6 and 16 kg, until 150 ms after box lift-off. Moreover, lifting of the overestimated load mass caused a disturbance of balance in 92% of the trials. The postural reactions aimed at regaining balance were not accompanied by an increased low-back loading. It was concluded that the absence of load knowledge, and the following overestimation of the load mass to be lifted, lead to an increased mechanical load on the lumbar spine and to an increased risk of losing balance in lifting tasks. Both events may contribute to a higher risk of low-back injury in manual materials handling tasks.

Introduction

In the industrialized world, the high prevalence of back injuries has developed into a well-recognized health problem. The associated costs in terms of loss of productivity, health care and individual suffering are unacceptably high. The life-time prevalence of low-back pain is estimated to be between 55% and 87% (Nicolaisen and Jorgensen 1985; Riihimäki 1985; Videman et al. 1984). Manual materials handling, especially lifting loads, is associated with low-back injuries (Andersson 1981; Chaffin and Park 1973; Klein et al. 1984). The high mechanical stress on the low-back involved is assumed to be a major cause (Frymoyer and Pope 1978; Nachemson 1978). Falls, slips and trips that occur while lifting a load, are also associated with low-back injuries (Manning and Shannon 1981). It is noteworthy that back injuries caused by falls are followed by longer sickness-absence and a higher rate of recurrence than accidental back injuries with other causes (Troup et al. 1981).

In many jobs, for example refuse collecting and luggage dispatching, the daily routine involves manual handling of loads with unknown mass. The one year prevalence of musculoskeletal complaints in the low-back region was 45% in refuse collectors (Stassen et al. 1993). Compared with other civil service workers, refuse collectors were more often declared unfit for work due to musculoskeletal injuries and neurological back problems (Verbeek and Geurts 1987). It might be hypothesized that the absence of load knowledge is a factor that contributes to the high prevalence of low-back injuries in refuse collectors. In lifting a load of unknown weight, the lifter may overestimate the weight and apply a larger force than is required to displace the actual load. As a result, the acceleration of lifter and load will be greater than expected and the lifter will move upward rapidly in an uncontrolled manner (Butler et al. 1993; Patterson et al. 1987). In this case, three adverse effects of the lack of load knowledge can be put forward. In the first place, the lifter can fall and strike the (low) back against an object or the floor. One per cent of the lumbo-sacral injuries reported in a gearbox factory was attributed to this event (Manning and Shannon 1981). In the second place, the overestimate of the load mass will cause greater forces at the hands and, consequently, at the low-back than actually needed for that particular lift. Experimental studies that investigated the effect of load knowledge on low-back loading, found that lifting an object with versus without load knowledge resulted in an increased lumbo-sacral loading in the latter condition (Butler et al. 1993; Patterson et al. 1987). Thirdly, the postural reactions required to regain balance could be hazardous to the low-back musculoskeletal system, as suggested by Oddsson (1990). Epidemiological studies have indeed shown that workers exerting sudden unexpected maximal efforts are particularly vulnerable to low-back disorders (Magora 1973).

Efforts to reduce the incidence of low-back pain at the workplace are often based upon the evaluation of manual materials handling tasks, in which several load determining factors are involved, e.g. the load location, the displacement of the load, the asymmetry of lifting, the lifting frequency, the coupling between load and hands. The NIOSH equation provides a method for computing a weight limit for manual lifting from these factors (Waters et al. 1993). The (absence of) load knowledge, however, is a factor that is not accounted for in this equation, nor in any other evaluation approach.

Load knowledge proved to be essential in lifting small objects with a precision grip, because the vertical lifting force pattern was found to be scaled to the object's weight (Forssberg et al. 1992). In bimanual lifting tasks involving the whole body, a similar scaling of the vertical force pattern can be assumed, for which adequate load knowledge would be required. Moreover, correct knowledge about the load mass to be lifted seemed important to make adequate preparations to counteract the threat to balance, that is imposed upon the lifter by picking up a load in front of the body (chapter 5).

The present study was aimed at gaining more insight into the effect of (the absence of) load knowledge on low-back loading and the control of balance in bimanual, whole-body lifting tasks. The mechanical load on the lumbar spine and the control of balance were studied in lifting tasks, in which subjects did have load knowledge and in tasks in which subjects did not always have the correct load knowledge. We first investigated how the lifter prepared himself to counteract the threat to balance, that is imposed by picking up a rather heavy load in front of the body. Next, we investigated whether the low-back loading was indeed increased in case subjects overestimated the load mass to be lifted and in case subjects showed postural reactions to prevent falling. Furthermore, the reasons for losing balance when subjects overestimated the load mass were studied.

Methods

Experiment I

Subjects and experimental procedures

Eight healthy male subjects (mean age 22.3 ± 1.5 (standard deviation) years, body height 1.79 ± 0.07 m, body mass 71 ± 11.7 kg, footlength 0.267 ± 0.067 m) participated in the experiment after they had given written informed consent and after approval of the Faculty's ethical committee. None of the subjects reported a history of low-back disorders or other motor impairments.

The subjects were asked to pick up and lift a barbell, in an ongoing down- and upward movement, and to come to a full stop holding the barbell at acromion height, using a leglift (straight back, bent legs) and a backlift (straight legs, bent back) (Figure 6.1). Several measures were taken to enhance the threat to balance that is imposed upon the lifter by picking up a load in front of the body and, thus, to enhance the necessity to make adequate preparations beforehand. (1) The barbell was fairly heavy, 22% of the subject's body mass. (2) The barbell was placed in front of the toes, at such a distance that the subject was just able to pick it up (heel-barbell distance 0.615 ± 0.054 m). This distance was similar for both lifting techniques. (3) In a series of ten to fifteen trials (for each technique condition), the lifting speed was increased each next trial, until the subject was no longer able to preserve balance (the duration of one downward movement phase thus decreased from circa 1.2 to circa 0.5 s). Imbalance was judged to occur when the heels lost contact with the ground or when a compensatory step was made to prevent falling.

Lifting speed was imposed by means of an acoustic metronome. Subjects first performed several movement cycles without picking up the load. When the required rhythm was attained, one of the authors counted down to the moment of barbell lift-off, starting at the beginning of the penultimate downward phase. To standardize the execution of the lifting tasks, the subjects were instructed to restrict their movements to the sagittal plane, to keep the heels on the ground at load pick-up and to guard their balance throughout the movement. In the lowest position, the vertical distance between the load and the ground was standardized at 14% body height. Subjects were instructed to lift the barbell in a straight vertical line, indicated by two flexible metal wands that were positioned in front of the barbell at the left and right end (from the subject's perspective). They performed practice trials to familiarize themselves with the tasks.

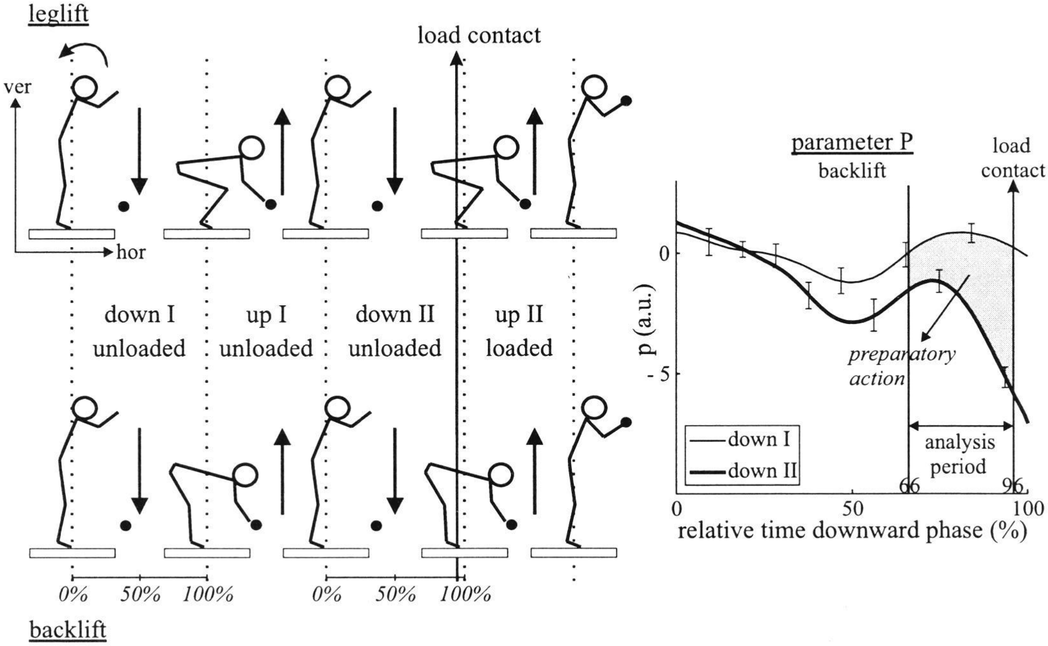


Figure 6.1
Two sequences of stick-figures illustrate the protocol of experiment I (left panel). Subjects lifted a barbell (up II) after several unloaded movement cycles, using a leglift (upper row) or a backlift (lower row). Two complete movement cycles were recorded (down I, up I, down II and up II). The beginning of a downward phase (at the highest body centre of mass position) was defined 0% relative time and the end (at the lowest body centre of mass position) was defined 100%. The moment of load contact occurred at 96% relative time. Positive directions for translation and rotation are shown in the upper left stick-figure. The right panel illustrates an example of a preparatory action (i.e. a significant difference between down II and I in the last 30% before load contact, filled area) in a (fictitious) parameter P for the backlift.

Linked segment model, kinematics and kinetics

The last two lowering-lifting cycles (Figure 6.1) were recorded using a 3-D semi-automatic video-based motion registration system (VICON™, Oxford Metrics Ltd.). Reflective markers (ø 25 mm) were attached to the skin (right side) to indicate the location of the fifth metatarsophalangeal joint, the ankle joint (distal part of the lateral malleolus), the knee joint (lateral epicondyle), the hip joint (greater trochanter), the lumbo-sacral (L5-S1) joint (as in

Looze et al. 1992), the spinous process of the first thoracic vertebra, the head (caput mandibula), the lateral border of the acromion, the elbow joint (lateral epicondyle), the wrist joint (ulnar styloid), and the hand (a small stick attached to the third metacarpus). An additional marker was attached to the right end of the barbell (from the subject's perspective). The coordinates of the acromion marker were used to determine the position of the shoulder joint. The length of the base of support was inferred from markers placed on the heel and the distal end of the most prominent toe. The coordinates of the joint positions defined eight body segments in the sagittal plane: the feet, lower legs, upper legs, pelvis, trunk/head, upper arms, forearms and hands(/load). The marker positions were sampled at 60 Hz and the raw sagittal plane coordinates were low-pass filtered with a digital filter (zero phase lag, 2nd order Butterworth, 5 Hz). Anthropometric data (body mass, length of segments) were measured. The mass of each segment, the positions of the segmental centres of mass (CoM), except for the trunk, and the moments of inertia were calculated according to Plagenhoef et al. (1983) and Looze et al. (1992). The mass, inertia and location of the CoM of the hands were adapted at the instant the hands grasped the load. The coordinates of the markers on the spinous process of the first thoracic vertebra and L5-S1 joint were used to determine the position of the trunk CoM during the movement according to an optimization procedure, which improved the estimated trajectory of the body CoM (Kingma et al. 1995). The angles of each segment were calculated relative to the right horizontal. Numerical differentiation (Lanczos 5-point differentiation filter) of the time histories of the segment angles and CoM positions yielded (angular) velocities and accelerations.

The ground reaction force was recorded by means of a strain gauge force platform (1.0 by 1.0 m). The analog force signals were amplified, low-pass filtered (30 Hz, 4th order), sampled (60 Hz, 12 bits) and stored synchronously to the movement registration by the VICON-system. For translations, the caudo-cranial and dorso-ventral directions were defined positive and for rotations the counter-clockwise direction (Figure 6.1, upper left stick-figure).

Biomechanical analysis

To obtain an estimate of the mechanical load on the lumbar spine, the net sagittal plane torque at the lumbo-sacral joint ($\boldsymbol{T}_{L5\text{-}S1}$) was determined by means of inverse dynamic analysis (Elftman 1939) using the dynamic 2-D linked segment model described by Looze et al. (1992). To study the control of balance, the position of the CoM relative to the base of support (CoM_{rel}), the horizontal and vertical momentum of the centre of mass ($\boldsymbol{p}_{hor}$, $\boldsymbol{p}_{ver}$)

and the angular momentum of the whole body (***L***) were determined. In a static situation, like upright standing, the CoM_{rel} has to remain within the borders of the base of support (Massion 1992). In a dynamic situation, like load lifting, other criteria for maintaining balance are operative such as confining the linear and angular momenta to certain limits during task execution (Toussaint et al. 1995). The parameters were determined as follows, according to Toussaint et al. (1995):

- CoM_{rel}: the horizontal position of the CoM of the body (including the load after pick-up), expressed as a percentage of the base of support (heels at 0% and toes at 100% CoM_{rel}).
- $\boldsymbol{p}_{hor}$ and $\boldsymbol{p}_{ver}$: the instantaneous horizontal and vertical momentum of the CoM of the body (including the load after pick-up), calculated from respectively the sum of the horizontal and the sum of the vertical momenta of all segmental CoMs.
- ***L***: the instantaneous angular momentum of the body (including the load after pick-up) around a rotation axis situated in the body's CoM, calculated from the sum of the segmental angular momenta.

Data analysis and statistics

The duration of all individual curves of a downward phase was normalized to 100% (Figure 6.1, left panel). The moment of barbell contact was defined four samples (67 ms) before the vertical displacement of the barbell exceeded 2.5 mm. The 194 successfully recorded trials were marked by one of the authors as 'balance' (115) or 'imbalance' (79), according to the criterium mentioned above. The latter category was not further analysed. To identify any specific preparations, kinematics and kinetics of the last downward phase (down II, Figure 6.1), before the barbell was lifted, were compared with the same parameters of the previous downward phase (down I, Figure 6.1), after which no load was lifted. A difference between both phases would be indicative of preparatory actions, of which an example is shown in the right panel of Figure 6.1. Before load contact, the values of parameter P in downward phase II decrease relative to the values of downward phase I. A significant difference between down II and I in the analysis period (see below) is considered a preparatory action.

A repeated measures multivariate analysis of covariance (MANCOVA) was performed on the five biomechanical parameters (dependent variables), with downward phase (down I, down II) and lifting technique (leglift, backlift) as within-subject factors, with subject (1 to 8) as between-subject factor and the average speed of the body CoM in both downward phases as a covariate. From the 115 balance trials, 40 leglift and 40 backlift trials were selected, such that each leglift-backlift pair was from the same subject, yielding 40 cases with 2^2 independent variable levels per case. For each parameter and each independent variable level, the last 30% of the downward phase *before* barbell contact (analysis period, Figure 6.1) was averaged and tested in the MANCOVA. This period was arbitrarily chosen, because testing at one point in time might fail to reveal a main effect, whereas analysing a period longer than 30% might average existing parameter changes too much. For each parameter, univariate F-tests were applied to interpret the overall effects. Effects were considered to be significant at $p<0.05$.

Experiment II

Subjects and experimental procedures

Twenty-five healthy male subjects (age 22.8 ± 2.0 years, body height 1.78 ± 0.03 m, body mass 73.4 ± 9.9 kg) participated in this experiment. None of them reported a history of low-back disorders or other motor impairments. All subjects were informed that they had to perform a series of tasks, in which a box of which the mass ranged from 6 to 16 kg was to be lifted. They were not informed about the sudden changes in load mass that were going to take place. The Faculty's ethical committee approved of the experimental set-up and judged the risk of falling backward involved with overestimating the load to be lifted acceptable. In a pilot study we established that an actual fall never occurred because of adequate postural reactions. After the experiment, the subjects were kindly requested not to reveal the experimental set-up to others.

Subjects were induced to overestimate the weight to be lifted, because they first lifted a box of 16 kg for two, three or four times (randomly assigned) and then lifted a box of 6 kg, only once. To make sure that subjects would expect to lift the 16 kg box instead of a 6 kg box, two black PVC boxes of equal size (0.24 x 0.34 x 0.42 m) and color, but different mass (6 kg v. 16 kg) were used. Upon completion of each lift, the subject was instructed to turn around. One of the authors removed the box and replaced the same or another box after 30 s. In this way, the expectation pattern of lifting 16 kg boxes was suddenly disrupted, leading to an overestimation of the 6 kg box. Imbalance was judged to occur when the forefoot lost contact with the ground or when a compensatory step was made to prevent falling. Subjects were instructed to lift as quickly as possible to prevent them from perceiving the actual load mass in the initial part of the lift. The duration of the upward movement phase was about

1 s. The subjects were standing in front of a box and upon a sign of one the authors flexed forward, grasped the box and lifted it to return to an upright position with the box held aloft at chest height. The CoM of the box, indicated by three reflective markers on the right side of each box, was placed 0.30 m in front of the subject's toes. The subjects were instructed to keep the heels on the ground at load pick-up, to restrict their movements to the sagittal plane and to guard their balance throughout the movement. No specific instructions were given regarding lifting technique. The subjects performed practice trials using the 16 kg box to familiarize themselves with the experimental task. Each subject performed at least two series of lifts, separated by a 15-min pause. Although most subjects expected the sudden change in weight on the second series, the strict instructions, high lifting speed and unexpectedness of the change constrained them to perform the second series just as the first one.

Linked segment model, kinematics and kinetics

The kinematics, kinetics and linked segment model of experiment II were similar to those of experiment I. For each trial, data collection started circa 0.5 s prior to the start of the lift and ended at the moment the subject was standing completely erect again.

Biomechanical analysis

The biomechanical analysis of experiment II was completely similar to that of I.

Data analysis and statistics

To permit averaging of trials, each trial was synchronized in time to the moment (t=0) the displacement of the CoM of the box exceeded 5 mm (box lift-off). A total of 81 samples was taken into analysis, 20 before box lift-off and 60 thereafter. An imbalance trial, resulting from lifting the overestimated load mass, was matched with the balance trial preceding it, yielding two experimental conditions: '16 kg' (balance) and '6 kg' (imbalance). To examine the effect of weight overestimation on the low-back loading and the control of balance, the biomechanical parameters of lifting the 16 kg box were compared with those of lifting the 6 kg box, that was expected to be 16 kg. A multivariate analysis of variance (MANOVA) was performed on the complete time traces of the five biomechanical parameters, with 50 cases and with experimental condition (16 kg, 6 kg) and time (81 samples) as between-subject factors. Univariate F-tests were applied to interpret the overall effects and effects were considered to be significant in case of p<0.05.

Results

Experiment I

Figure 6.2 shows the five parameters studied in relation to the preparatory actions which a lifter performs before picking up and lifting a load. In each panel four time traces are displayed, representing different combinations of downward phase (I, II) and lifting technique (leg-, backlift). The pick-up of the load induces a quick forward shift of the projection of the CoM on the ground (CoM_{rel}). The upper right panel depicts a part of this forward shift: from load contact onwards (96%) the time traces of down II increased quickly for both lifting techniques, i.e. the CoM_{rel} moved towards the toes. This is a balance threatening event, because the CoM approached the front margin of the base of support. Furthermore, picking up the load in front of the body brakes the counter-clockwise (positive) angular momentum $\boldsymbol{L}$ of the body towards an erect standing posture (not visible in Figure 6.2). Since $\boldsymbol{L}$ was only small at that time (about zero at the end of downward phase I, upper left panel), load pick-up induced a risk of toppling forward.

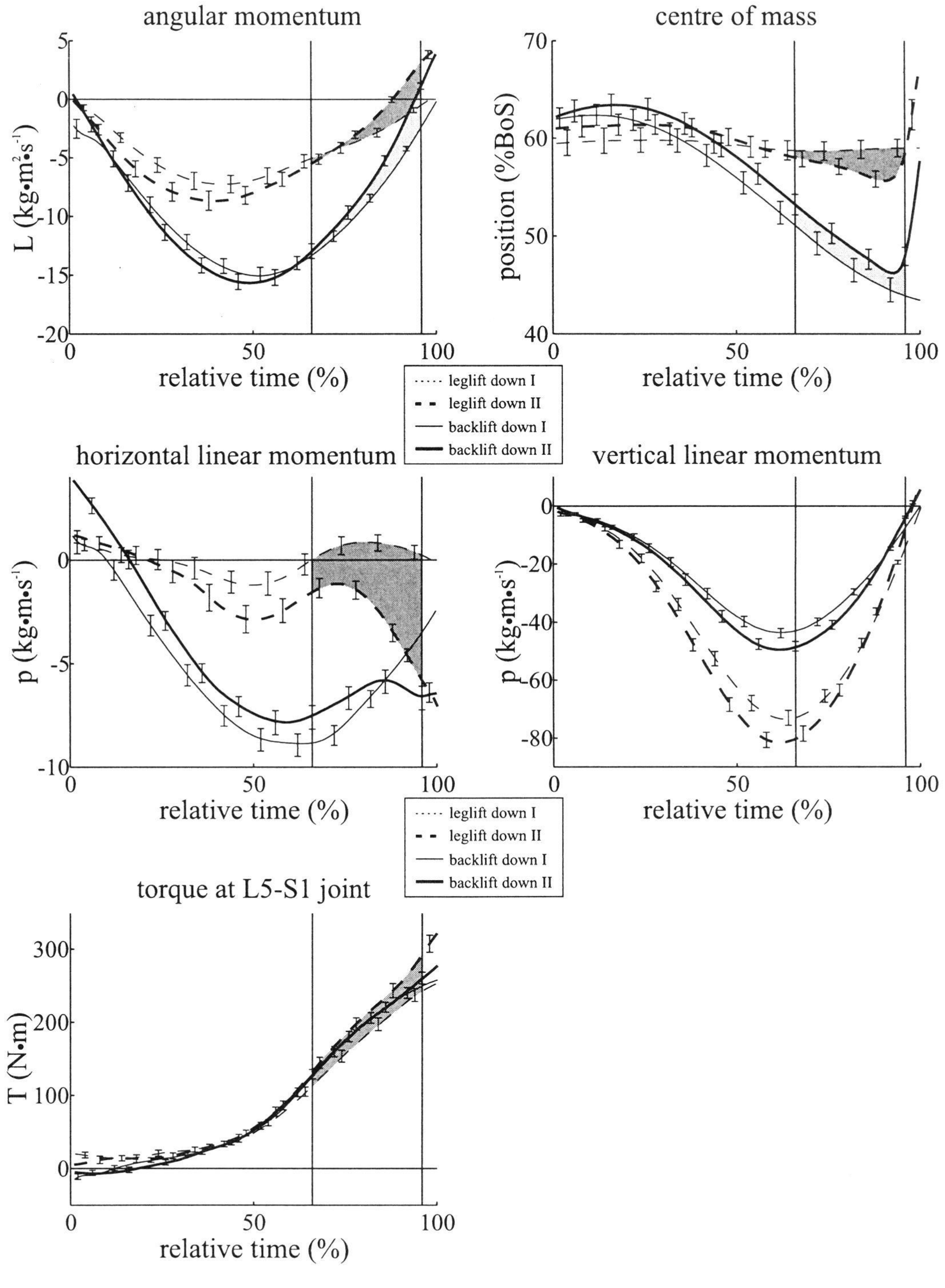
To counteract these threats to balance, preparatory actions were performed. For the leglift, the CoM_{rel} was positioned more towards the heels at the end of down II compared with I and $\boldsymbol{p}_{hor}$ (middle left panel) decreased considerably. For both techniques $\boldsymbol{L}$ was larger at the end of downward phase II, attaining positive values before load contact. Parameter $\boldsymbol{p}_{ver}$, however, did not display a preparatory action (middle right panel) and $\boldsymbol{T}_{L5\text{-}S1}$ (lower panel) was larger at the end of downward phase II for the leglift only. As can be deduced from the presence or absence of filled areas in Figure 6.2, the preparatory actions were not always similar for the two lifting techniques. For instance, in leglifting the CoM_{rel} was positioned more towards the heels in down II, compared with I, whereas in backlifting it was positioned a little more towards the toes in down II.

The differences between both downward phases were found to be significant (MANCOVA, Wilk's lambda 0.131, F=37.249, p<0.05), indicating that subjects prepared themselves before picking up the load. The preparatory actions were not the same for both lifting techniques, because a significant interaction effect between downward phase and technique was found (Wilk's lambda 0.301, F=13.014, p<0.05). Table 6.1 summarizes the univariate test results, presenting the differences between both downward phases for each lifting technique.

Table 6.1
Differences between downward phase I and II for five biomechanical parameters, presented for the two lifting techniques separately. The signs '↑' and '↓' represent, respectively, a significant increase and decrease in the mean parameter value during the analysis period from downward phase I to II, while '0' indicates that no significant difference was found. * denotes a significant difference in preparation between techniques.

Parameter		Lifting technique		
		leglift (*n=40*)	backlift (*n=40*)	significant difference
$\boldsymbol{p}_{hor}$	($kg{\bullet}m{\bullet}s^{-1}$)	↓	0	*
$\boldsymbol{p}_{ver}$	($kg{\bullet}m{\bullet}s^{-1}$)	0	0	
CoM_{rel}	(%BoS)	↓	↑	*
$\boldsymbol{L}$	($kg{\bullet}m^2{\bullet}s^{-1}$)	↑	↑	
$\boldsymbol{T}_{L5\text{-}S1}$	($N{\bullet}m$)	↑	0	*

Figure 6.2
Time traces of two downward phases (I and II) of a lifting task for the angular momentum (upper left), the horizontal position of the centre of mass relative to the base of support (BoS) (upper right), the horizontal and vertical linear momentum (middle panels) and the lumbo-sacral (L5-S1) torque (lower panel) for the leglift and backlift. Mean time trace ± 1 standard error of the mean (SEM) (*n=40*) are shown. Conventions as in Figure 6.1, with the dark gray area marking a preparatory action during leglifting and the light gray area during backlifting.


angular momentum
L (kg•m²s⁻¹)
relative time (%)
centre of mass
position (%BoS)
relative time (%)
leglift down I
leglift down II
backlift down I
backlift down II
horizontal linear momentum
p (kg•m•s⁻¹)
relative time (%)
vertical linear momentum
p (kg•m•s⁻¹)
relative time (%)
leglift down I
leglift down II
backlift down I
backlift down II
torque at L5-S1 joint
T (N•m)
relative time (%)

Experiment II

Movement characteristics of the lifting task

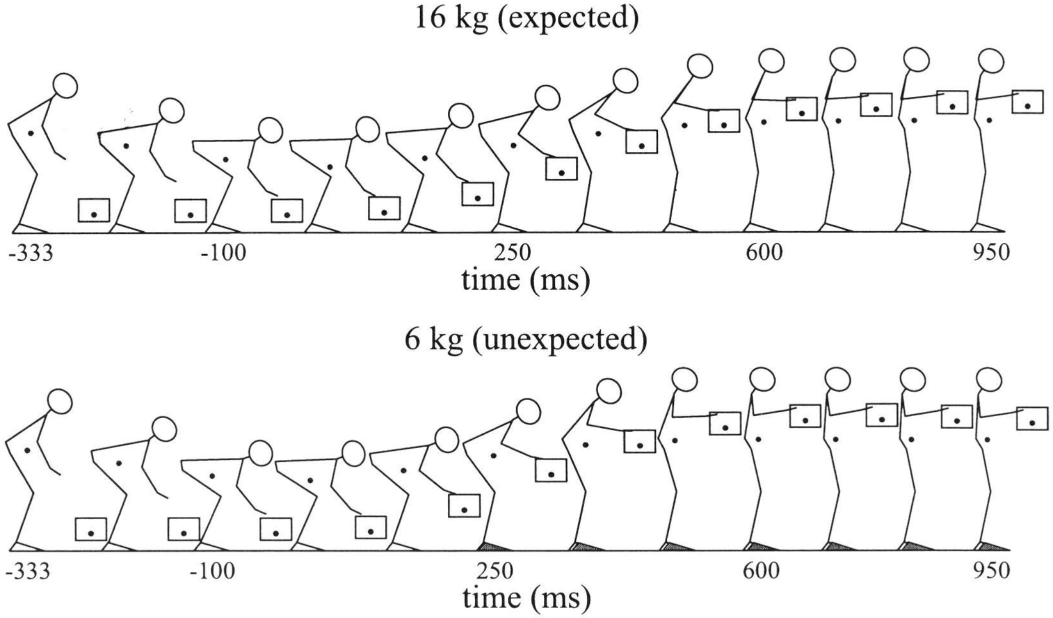


Figure 6.3
Two sequences of stick-figures represent the average movement of subjects during fifty 'expected 16 kg' trials (upper row) and fifty matched 'unexpected 6 kg' trials (lower row). The centre of mass of the whole body and of the box are indicated by dots. The first stick-figure is at 20 samples (333 ms) before time 0, which was the moment at which the vertical displacement of the box's centre of mass exceeded 5 mm. The time interval between two adjacent stick-figures is 7 samples (117 ms). In the last seven stick-figures of the lower row, the (additional) shaded foot segment displays the foot position at t=250 ms to indicate that some subjects made a backward step.

The attempt to wilfully induce a loss of balance by creating an overestimation of the object's weight was successful in the second experiment: in 92% of the lifting trials, in which the box's weight was 6 instead of 16 kg, subjects showed signs of compensatory responses such as lifting of the forefoot or making a backward step to prevent falling. Two sequences of stick-figures in Figure 6.3 represent the average movement performed during the last part of the unloaded downward and the complete loaded upward phase for both experimental conditions. The subject picked up the box in a continuous motion and lifted it in a vertical line to chest height. The stick-figures of the 16 kg trials closely resemble those of the 6 kg trials, especially before the box is grasped. However, some differences are noticeable after box lift-off: from 250 to 717 ms the subjects extended their backs faster in the 6 kg trials than in the 16 kg trials. Some compensatory movements are visible from 250 ms onwards: subjects held the box further in front of the body and made a backward step, indicated by the backward shifted foot-segment.

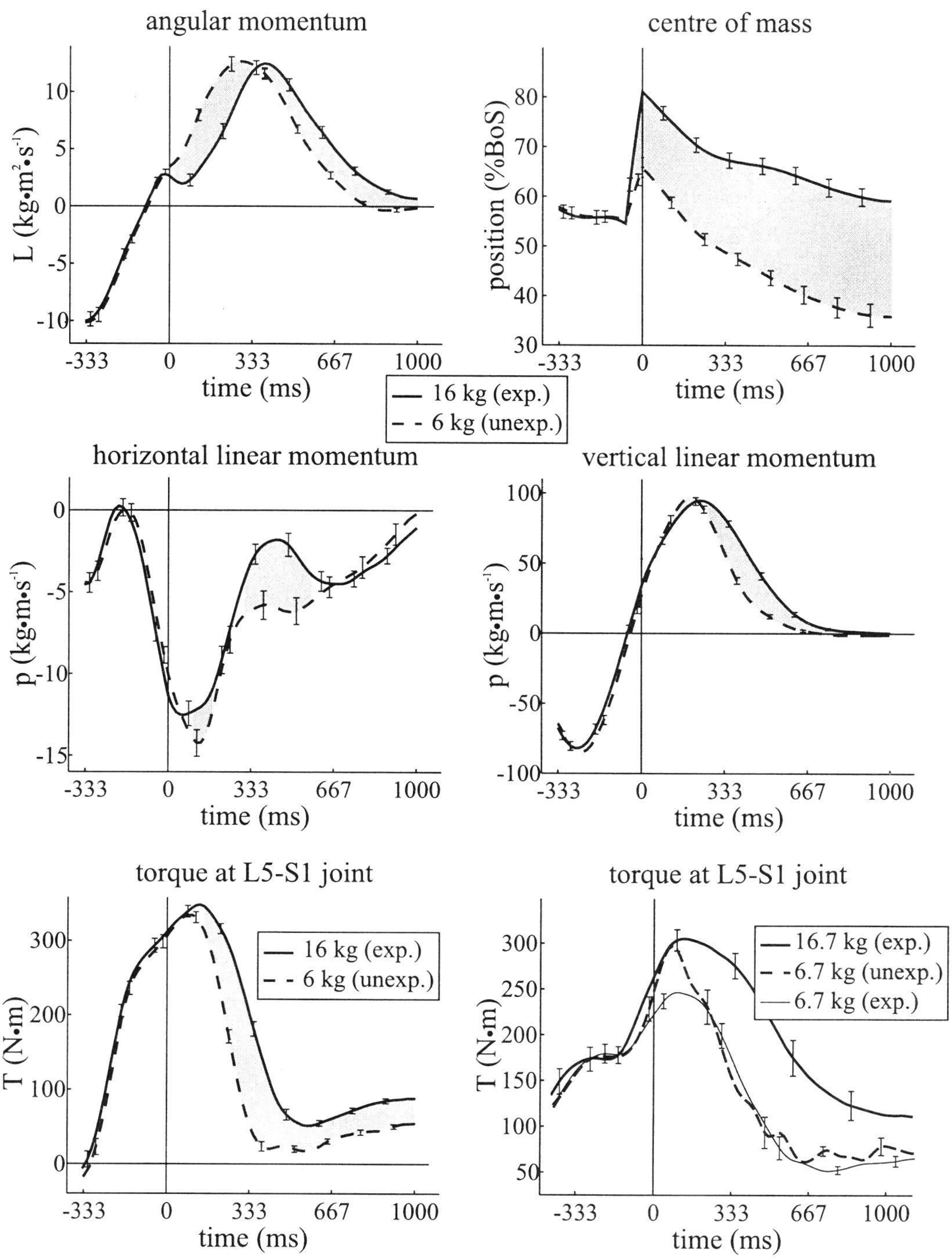
The effect of overestimating the load mass on low-back loading and control of balance

When unexpectedly presented with the 6 kg box, subjects did not prepare themselves to lift the 6 kg box, but the expected 16 kg box. Thus, biomechanical parameters should be the same prior to lifting the 16 kg versus the 6 kg box. Furthermore, the biomechanical parameters should picture the signs of imbalance when lifting the overestimated load mass. Figure 6.4 shows the time traces of the five biomechanical parameters for both experimental conditions, from 333 ms before box lift-off until 1000 ms thereafter. Figure 6.4 clearly shows no difference between the time traces of both experimental conditions prior to grasping the box. This finding was confirmed by the results of the MANOVA, performed on the times traces of the five biomechanical parameters. A significant interaction effect of experimental condition and time was found (Wilk's lambda 0.548, F=12.665, p<0.05), indicating that the time traces of lifting the 16 kg box were indeed different from those of lifting the 6 kg box, but not during the whole time period analysed. Periods of significant differences between the two conditions are filled in the graphs. The univariate analyses revealed a significant difference between both conditions around box lift-off for parameters $\boldsymbol{L}$ and $\mathrm{CoM}_{\mathrm{rel}}$ (upper panels), at +17 ms and -33 ms respectively. For $\boldsymbol{p}_{\mathrm{hor}}$ and $\boldsymbol{p}_{\mathrm{ver}}$ (middle panels) the deviation started longer after box lift-off, at 100 ms in both cases. The finding that $\boldsymbol{T}_{\mathrm{L5\text{-}S1}}$ (lower left panel) was different between both conditions first after 150 ms implies that the peak low-back loading was similar when lifting a 16 kg box and when lifting a 6 kg box. Thus, the absence of load knowledge, and the subjects' assumptions that the box was 16 instead of 6 kg, resulted in a low-back loading that would accompany the lifting of a 16 kg box. Only after box lift-off the subjects sensed that the box was lighter than 16 kg and gradually the low-back loading became less than the low-back loading of lifting a 16 kg box. However, this does not confirm the hypothesis that the overestimation of the load mass to be lifted resulted in an increased mechanical load on the lumbar spine. Therefore, we repeated this experiment and extended the protocol: after lifting of the overestimated load mass, the subjects lifted that box for another three times while they knew that the load mass was reduced. Eight (other) healthy male subjects lifted a box, that was provided with force transducers in the handles, using a freely chosen lifting technique and speed. They first lifted a 16.7 kg box four times, then lifted a 6.7 kg box which they expected to be 16.7 kg and finally lifted this 6.7 kg box three times knowing its actual weight. Per subject, one series of eight trials was performed. In the fifth trial, all subjects indeed overestimated the load mass to be lifted and they had to make compensatory responses after box lift-off. The lower right panel of Figure 6.4 again shows that $\boldsymbol{T}_{L5\text{-}S1}$ was not different in the initial part of the upward phase when lifting the 'expected' 16.7 kg box and the 'unexpected' 6.7 kg box; peak $\boldsymbol{T}_{L5\text{-}S1}$ was not significantly different (309.75 v. 303.29 N•m, t=0.92, p=0.387). However, lifting of the 'unexpected' 6.7 kg box did significantly increase $\boldsymbol{T}_{L5\text{-}S1}$ during the first period after lift-off when compared with lifting of the 'expected' 6.7 kg box (peak $\boldsymbol{T}_{L5\text{-}S1}$ 303.29 v. 248.82 N•m, t=-5.90, p<0.05). From circa 175 ms after lift-off onwards, the $\boldsymbol{T}_{L5\text{-}S1}$ curves in both 6.7 kg conditions were similar, suggesting that the subjects had adjusted their movement pattern and force generation on the box to the actual weight.

Figure 6.4 also shows what happened with respect to the control of balance. Overestimating the load mass caused an 'overshoot' in the linear momenta. Just after box lift-off, the negative (backward) $\boldsymbol{p}_{hor}$ reached a larger value in the 6 kg trials than in the 16 kg trials. Likewise, the positive (upward) $\boldsymbol{p}_{ver}$ increased more when the 6 kg box was lifted. The positive (counter-clockwise) $\boldsymbol{L}$ showed an overshoot too when the overestimated load mass was lifted. Thus, the linear and angular momenta were larger than expected, leading to imbalance and compensatory reactions to regain balance. These reactions were not accompanied by a higher low-back loading; the 6 kg $\boldsymbol{T}_{L5\text{-}S1}$ did not show an overshoot.

Figure 6.4

Time traces of two different lifting tasks (lifting 16 kg v. lifting 6 kg that was expected to be 16 kg) for the angular momentum (upper left), the horizontal position of the centre of mass relative to the base of support (BoS) (upper right), the horizontal and vertical linear momentum (middle panels) and the lumbo-sacral (L5-S1) torque (lower left panel). Mean time trace ± 1 SEM (n=50) are shown. Time 0 indicates box lift-off, the moment at which the vertical displacement of the box's centre of mass exceeded 5 mm Periods of significant differences between the two experimental conditions are filled. The lower right panel presents the lumbo-sacral (L5-S1) torque of an additional experiment. Mean time traces ± 1 SEM (n=8) of three different lifting tasks (16.7 kg expected, 6.7 kg that was expected to be 16.7 kg and 6.7 kg expected) are shown. Time 0 indicates box lift-off, the moment at which the vertical force applied on the box exceeded the weight of the box. Note that the performance time of these tasks was considerably longer than the time of the tasks in the lower left panel, which explains the higher peak torques in that graph.


angular momentum
L (kg•m²•s⁻¹)
time (ms)
centre of mass
position (%BoS)
time (ms)
16 kg (exp.)
6 kg (unexp.)
horizontal linear momentum
p (kg•m•s⁻¹)
time (ms)
vertical linear momentum
p (kg•m•s⁻¹)
time (ms)
torque at L5-S1 joint
T (N•m)
time (ms)
16 kg (exp.)
6 kg (unexp.)
torque at L5-S1 joint
T (N•m)
time (ms)
16.7 kg (exp.)
6.7 kg (unexp.)
6.7 kg (exp.)

Discussion

Maintaining balance when picking up a load of known mass

Picking up a load in front of the body induced a risk of toppling forward, because the body's CoM quickly shifted forward and the counter-clockwise angular momentum of the body towards an erect posture was braked. Hence, the projection of the CoM on the ground approached the front margin of the base of support and a smooth extending movement of the subject was hampered. Apparently, the subjects successfully minimized the adverse effects of these balance threatening events, since balance was not lost. This was accomplished by specific preparations prior to load pick-up, demonstrated in experiment I. A discussion of the preparations from a motor control point of view can be found in chapter 5 and Toussaint et al. (in press 1997a; in press 1997b).

In the first place, the adverse effect of the forward CoM shift was reduced by the preparatory change in the CoM momentum $\boldsymbol{p}_{\mathrm{hor}}$. During leglifting, a profound backward $\boldsymbol{p}_{\mathrm{hor}}$ was created prior to load pick-up, to brake the forward CoM shift and thus prevent that the $\mathrm{CoM}_{\mathrm{rel}}$ approached or even crossed the front margin of the base of support after load pick-up. During backlifting, a decrease in the backward directed $\boldsymbol{p}_{\mathrm{hor}}$ occurred close to load pick-up, but it was not significant. Without preparation (see down I in Figure 6.2), the backward $\boldsymbol{p}_{\mathrm{hor}}$ would have been smaller or $\boldsymbol{p}_{\mathrm{hor}}$ would even have been directed forward. In the second place, the adverse effect of a braked counter-clockwise angular body momentum $\boldsymbol{L}$ was reduced by a preparatory increase in $\boldsymbol{L}$ for both techniques. Without the preparatory increase in $\boldsymbol{L}$ (see down I in Figure 6.2), $\boldsymbol{L}$ would probably be too small at load pick-up and the counter-clockwise rotation might then even be reversed to a clockwise rotation, which could induce a fall forward. In short, loss of balance when picking up a load was prevented by specific preparations that yielded a large backward CoM momentum and a considerable counter-clockwise angular momentum at load pick-up. It is important to remember that load knowledge was required to execute these preparations, for absence of load knowledge yielded inadequate preparations in experiment II.

Control of balance and lifting technique

The preparatory changes described above were not exactly the same for the leglift and the backlift (Figure 6.2 and Table 6.1). This is interesting, because a difference in preparatory actions implies that the balance threatening effect of load pick-up was not the same in both techniques. Although the backlift and leglift have been often subjected to research to identify and study differences in, for instance, low-back loading (Toussaint et al. 1992), metabolic energy expenditure (Looze et al. 1992b) or spinal shrinkage (Dieën et al. 1994), differences in control of balance have never been assessed.

A significant difference in preparation between techniques was found for $\boldsymbol{p}_{\text{hor}}$, suggesting a differential effect of load pick-up on the body's CoM position. Without preparation (down I), the CoM_{rel} was positioned closer to the front margin of the base of support in leglifting compared with backlifting. The forward CoM shift at load pick-up was, therefore, more threatening when leglifting. Furthermore, a technique difference in the direction and magnitude of $\boldsymbol{p}_{\text{hor}}$ at load pick-up was observed. Without preparation, the leglift $\boldsymbol{p}_{\text{hor}}$ was about zero at load pick-up, while the backlift $\boldsymbol{p}_{\text{hor}}$ was negative, that is, directed backward. Hence, in backlifting, the forward CoM shift at load pick-up would be reversed by the backward $\boldsymbol{p}_{\text{hor}}$, even without preparatory actions, whereas in leglifting the CoM would shift forward without being braked or reversed. Thus, these results suggest that preserving balance while picking up a load with a backlift was easier than while picking up the same load using a leglift. This suggestion is supported by the observation that in 44% of the leglift trials a loss of balance was observed, whereas this occurred in 36% of the backlift trials.

Absence of load knowledge and low-back loading

Epidemiological studies have demonstrated that part of the low-back injuries in industry is associated with falls, slips or trips that occur while lifting a load (Manning and Shannon 1981; Troup et al. 1981). Experiment II demonstrated that absence of load knowledge could lead to balance loss when the load mass was overestimated and, thus, may explain a part of the low-back pain incidence. The question is whether the mechanical load on the lumbar spine was indeed increased in that case and, if so, whether the increase occurred in the preparatory phase or during the execution of postural reactions to regain balance.

The results of experiment II and the additional experiment demonstrated that the mechanical load on the lumbar spine was indeed increased when lifting an object without correct load knowledge, that is when the weight of the box was overestimated with 10 kg. Since subjects prepared themselves according to the expected load mass, no difference in low-back loading was observed between lifting a load of 16 kg and lifting a load of 6 kg that was expected to be 16 kg, until 150 ms after box lift-off (Figure 6.4). Even the peak low-back loading was similar in both cases, although the actual mass difference was 10 kg. The peak low-back loading was, however, significantly lower when subjects lifted the 6 kg box with the correct load knowledge. Thus, the *expected* load mass largely determined the peak low-back loading, rather than the *actual* mass. This finding has important implications for existing guidelines for safe low-back loading limits. The NIOSH equation, for example, provides a method for computing weight limits for manual lifting based on the actual load mass and several factors that determine the functional load on the low-back (Waters et al. 1993). The equation does not account for the expected load mass, which seems of more importance than the actual mass.

The postural reactions to prevent falling did not seem to affect low-back loading. Postural reactions presumably started 100 to 150 ms after box lift-off, as may be deduced from the sharp switch in the negative $\boldsymbol{p}_{\mathrm{hor}}$ of the 6 kg trials (Figure 6.4). Around that time, the low-back loading reached its peak and $\boldsymbol{T}_{\mathrm{L5\text{-}S1}}$ of the overestimated 6 kg trials started to decrease with respect to $\boldsymbol{T}_{\mathrm{L5\text{-}S1}}$ of the 16 kg trials.

Absence of load knowledge and control of balance

In 92% of all lifting trials in which subjects were induced to overestimate the box's weight, they lost balance, that is, they had to make serious efforts to prevent falling. Figure 6.4 elucidates what happened. Overestimating the load mass caused an overshoot in the linear and angular momenta; in the 6 kg trials, the back- and upward CoM momentum reached larger values than in the 16 kg trials. Thus, the subjects moved much faster back- and upward after box lift-off than intended, leading to imbalance. The existence of an overshoot in linear and angular momenta implies that the preparations in linear and angular momenta were not correct in the 6 kg trials. They were programmed to counteract the balance-disturbing effect of picking up a 16 kg load and, thus, were too large to match the effect of picking up a 6 kg load. Further details about adequately and erroneously programmed preparatory motor commands can be found in Toussaint et al. (submitted). Thus, experiment II clearly demonstrated that the preparatory actions, aimed at minimizing the balance-threatening effect of load pick-up, were programmed according to the *expected* load mass, not according to the (unknown) *actual* load mass. It proves that load knowledge is essential for adequate programming of the preparations.

Conclusions and practical implications

In industrial whole-body lifting tasks, workers presumably prepare themselves also before they pick up a load. These preparations are directed at minimizing the risk of losing balance that is inherent in picking up a load in front of the body. They are based upon the expected load mass and are specific for the applied lifting technique. The preparations also determine the peak mechanical load on the lumbar spine. In case similar objects are continuously handled (e.g. at an assembly line), the preparations will be programmed on the basis of the weight of the previous object. Such a preparation will not be correct, however, when a similarly looking object of different weight is suddenly presented (the expectation pattern is disrupted). When the lifter overestimates the object's weight, an overshoot in the linear and angular momenta of the body will occur, leading to a disturbed balance, to twisting or jerking actions to regain balance and possibly to a fall. The low-back loading will resemble the low-back loading during lifting of the expected (overestimated) load mass, not of lifting the actual mass. In case the task comprises handling of objects varying in size and load mass, information about the weight of the previous object is not useful. The weight of each object will have to be estimated on the basis of its size, a common density and comparison with other objects (Gordon et al. 1991b).

If workers are required to use programmed preparations, for instance when they have to lift at a predetermined, high pace, it is important to provide them with adequate information about the actual load mass. This implies (1) that the actual load mass has to be clearly and unequivocally displayed on the object in case objects of varying size, weight and unpredictable density are handled; (2) that workers should be trained in performing preparations that are adequate for the depicted load mass, because experience is required to perform adequate preparations when only visual information is present and (3) that a pattern of expected load masses (for instance in an assembly line) should not be suddenly disrupted. With respect to guidelines for safe low-back loading limits, the expected load mass should be accounted for in situations without load knowledge, because the expected load mass seems to determine the low-back loading, rather than the actual mass.

Acknowledgements
The authors would like to thank Marco Hoozemans, Michiel Ober, Marije Faber and Yvonne Michies for their support in data acquisition and processing. Michiel de Looze is gratefully acknowledged for advise and for reviewing the manuscript.

Chapter 7

Is regulation of the anterior-posterior centre of mass position the leading principle in the organization of postural adjustments in a bimanual, whole-body lifting task?

Abstract

The present chapter investigated whether the regulation of the anterior-posterior position of the body centre of mass (CoM) with respect to the base of support can be regarded as the leading principle in the organization of postural adjustments in a bimanual, whole-body lifting task. The anticipatory postural adjustments counteracting an expected equilibrium perturbation and the postural responses compensating an unexpected perturbation were examined applying a global mechanical analysis of the task. Eight subjects lifted a box, in an ongoing down- and upward motion, from 0.14 m above the floor to chest height. The anticipatory adjustments were investigated by comparing the metrics of a loaded movement cycle with those of an unloaded one. The magnitude of the expected perturbation was varied by lifting a box of 6.7 kg and one of 16.7 kg. The compensatory postural responses were examined in a lifting trial, in which the subjects were induced to anticipate on lifting the 16.7 kg-box, while the 6.7 kg-box was presented. The global mechanical analysis of the postural adjustments suggested that regulating the anterior-posterior CoM position was not the leading cue in the organization of these adjustments. Instead, regulation of the moment exerted by the ground reaction force about the CoM appeared to take precedence, while the CoM positioning seemed to be subordinated to this. As the external moment regulates the whole-body angular momentum, it is proposed that control of the amount of rotation of body segments, in particular of the trunk/head segment, is the leading principle in the postural adjustments during lifting.

Introduction

Lifting a large object with two hands from the floor to waist or chest height is an often occurring activity in daily life. The main goal of this so-called *bimanual, whole-body lifting task* is to displace an object to a defined end position, preferably in a smooth and continuous motion. To achieve such dexterous and successful task performance is, however, by no means trivial. Picking up and lifting an object from the floor in an ongoing down- and upward body motion means that an extra mass is added to a moving multi-joint system. Hence, the object's inertia may, depending on the differences in velocity, mass and position between object and body, decelerate the upward, extending motion of the subject. Thus, the act of picking up and lifting a rather heavy object presents the subject with an expected disturbance of the lifting movement and, thereby, an expected disturbance of the whole-body's equilibrium.

In the previous four chapters, it was assumed that regulation of the horizontal[2] centre of mass (CoM) position was a crucial element in the control of equilibrium in the bimanual, whole-body lifting task. This assumption was based on the general view that maintenance of the horizontal position of the CoM with respect to the base of support against external and internal disturbances provides the central rule for the organization of postural adjustments in upper-body tasks performed in stance (Massion 1992; 1994). Furthermore, this rule was found operative in the organization of anticipatory (Nashner and Forssberg 1986) and compensatory (Nashner 1980) postural adjustments in a whole-body task like locomotion. The results presented in those four chapters seemed to support the assumption. Firstly, a synergy between the motions of upper body and lower limbs was found to confine the forward displacement of the body CoM during the downward movement towards the load (chapter 3). Secondly, when performing the lifting task on a base of support that was reduced to 0.092 m in the horizontal direction, the horizontal CoM trajectory was shown to be shifted backwards towards the middle of the new base of support (chapter 4). Thirdly, the backward CoM velocity was increased prior to grasping the object. This anticipatory postural adjustment appeared to counteract the forward shift of the combined CoM of body and load,

[2] *given the sagittal plane description of the lifting task, the horizontal direction refers to the anterior-posterior direction in this chapter*

that is inherent in the addition of an extra mass anterior to the body (i.e. load pick-up) (chapters 5 and 6).

However, some results suggested that regulating the horizontal CoM position did not always take precedence in the organization of the postural adjustments. Firstly, the perturbing forward CoM shift was not entirely counteracted prior to its occurrence, but also thereafter, suggesting that additional rules may be crucial in organizing the postural adjustments before load pick-up (Toussaint et al. in press 1997a). Secondly, the anticipatory postural adjustments in horizontal CoM velocity were found to depend on the technique applied to lift the load (i.e. back- v. leglift), intimating that technique-specific characteristics of the lifting task were also important (chapters 5 and 6). Thirdly, anticipatory postural adjustments were found in the whole-body angular momentum, a parameter that quantifies the amount of rotation of body segments (chapter 5). This finding stresses that performance of the whole-body lifting task can not be simplified to the translation of the CoM relative to the base of support. The task is a multi-joint movement, with pronounced rotations of body segments in space and relative to each other. Thus, the control of equilibrium may involve the regulation of not only horizontal CoM translation, but also body segment rotation and at certain stages of task execution, control of the first may be subordinated to control of the latter. This suggestion would be at variance, though, with the general view that the horizontal CoM position is the cue on which the organization of the postural adjustments is based (Massion 1994). Therefore, the present study (re-)investigated the postural adjustments in a bimanual, whole-body lifting task to identify the leading principle(s) in the organization of these adjustments.

The equilibrium disturbance in a whole-body lifting task will not only affect the position of the CoM with respect to the base of support. Rather, the ongoing whole-body motion, in terms of horizontal and angular velocity, will be influenced. Thus, analysis of both perturbation and postural adjustments should not be restricted to the horizontal CoM position with respect to the base of support. Firstly, the present study will incorporate the horizontal CoM momentum (see also Pai and Patton 1997). Secondly, given the existence of a considerable rotational component in the task, the whole-body angular momentum will be analysed. And thirdly, the perturbation will be evaluated in terms of its effect on horizontal and vertical CoM momenta and whole-body angular momentum. A box with instrumented handles was used to record the external (box reaction) force that might affect the ongoing whole-body motion.

The anticipatory postural adjustments counteracting an expected equilibrium perturbation and the postural responses compensating an unexpected perturbation were investigated, applying a global mechanical analysis of the task. Eight subjects lifted the box, in an ongoing down- and upward motion, from 0.14 m above the floor to chest height. The anticipatory postural adjustments were analysed by comparing the kinematics and kinetics of this loaded movement cycle with those of the preceding unloaded cycle. Differences between both conditions were termed *anticipatory* if they occurred prior to the onset of the upward force on the box. The magnitude of the expected perturbation was varied by lifting a box of 6.7 kg and one of 16.7 kg. The postural responses to an unexpected equilibrium perturbation were examined in a lifting trial, in which the subjects were induced to anticipate on lifting the 16.7 kg-box, while the 6.7 kg-box was presented. This paradigm was shown previously to provoke a disturbance of equilibrium and subsequent postural reactions (chapter 6).

Methods

Subjects

Eight healthy male subjects (mean age 23.5 ± 2.4 (standard deviation -SD-) years, body height 1.81 ± 0.04 m, body mass 71.9 ± 9.4 kg, shoelength 0.281 ± 0.010 m) participated in the experiments. None of the subjects reported a history of low-back disorders or other motor impairments. All subjects were informed that they had to perform a series of lifts, in which a box of 6.7 or 16.7 kg was to be lifted. They were not informed that between trials the 16.7 kg load was going to be reduced to 6.7 kg. The Faculty's ethical committee approved of the experimental set-up and judged the risk of falling backward involved with overestimating the load to be lifted to be acceptable. In a previous study (chapter 6) it was found that an actual fall never occurred because of adequate postural reactions. After the experiment, the subjects were kindly requested not to reveal the experimental set-up to others.

General procedure and experimental protocol

The subjects were instructed to grasp a box with two hands in the middle of an ongoing downward-upward movement and to lift it in a symmetric motion to chest height. They performed several unloaded down- and upward movements before they actually grasped and lifted the box. To indicate the moment of load pick-up, one of the authors counted down to this instant, starting at the beginning of the penultimate downward phase. The data acquisition started about 0.5 s prior to that moment and ended about 1 s after the box had reached the intended position. Thus, successively, one completely *unloaded movement cycle* and one movement cycle with an unloaded downward and a loaded upward phase (*loaded cycle*) were recorded (see the left panel of Figure 7.1).

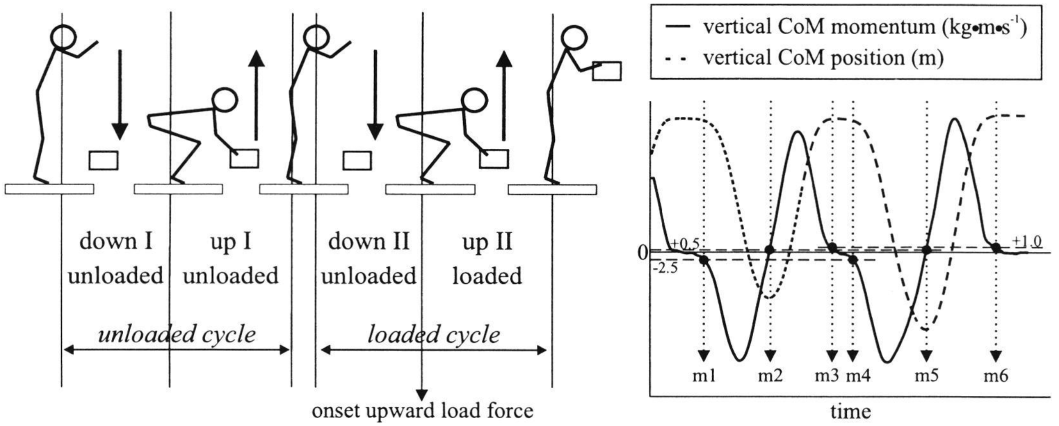


Figure 7.1
Schematic outline of the bimanual, whole-body lifting task and definition of the four movement phases. The vertical body centre of mass (CoM) position and momentum are time traces of a typical example. The vertical CoM position time trace was re-scaled (enlarged) and shifted (downwards). The abbreviations m1 to m6 are explained in the text.

No specific instructions were given regarding lifting technique, but all subjects performed the task by bending and extending both the legs and the lumbar spine. The box was placed in front of the subject at a standardized position; both the horizontal distance between the front of the feet and the back of the box handles and the vertical distance between the base of support and the bottom of the handles was 15% of the individual body height (0.271 ± 0.006 m). The subjects were instructed to apply the same speed of movement in each trial, to keep the heels on the ground during the whole task and to restrict their movements to the sagittal plane. The subjects first lifted the 6.7 kg-box, eight times in a row. Next, they were instructed to lift the same box to which, with their knowledge, an extra 10 kg was added another eight times. After each trial, the subjects were requested to step out of the experimental room. Without their knowledge, the extra 10 kg was removed from the box in the fifth trial. Thus, subjects expected to lift the 16.7 kg-box just like in the preceding four trials, whereas the actual mass was only 6.7 kg. Each trial was separated by a 1-min pause and a 5-min pause was allowed between the two load conditions. Before the experiment, the subjects performed about ten practice trials with the 6.7 kg-box to familiarize themselves with the task.

Apparatus, data acquisition and data processing

Kinematics

A 3-D semi-automatic video-based motion registration system (VICON™, Oxford Metrics Ltd.) recorded the positions of fourteen light-reflecting markers (ø 25 mm), with four cameras, at a sample rate of 60 Hz. Eleven markers were attached to the subject's right body side, to indicate the location of the fifth metatarsophalangeal joint (on the shoe), the ankle joint (distal part of lateral malleolus), the knee joint (lateral epicondyle), the hip joint (greater trochanter), the lumbo-sacral joint (L5-S1) (as in Looze et al. 1992), the spinous process of the first thoracic vertebra (Th1), the head (caput mandibula), the lateral border of the acromion, the elbow joint (lateral epicondyle), the wrist joint (ulnar styloid) and the hand (third metacarpophalangeal joint). Three markers were placed on the right side of the box. The coordinates of the acromion marker were used to determine the position of the shoulder joint. The raw sagittal plane coordinates of the markers were low-pass filtered with a digital filter (6 Hz, 2nd order Butterworth, zero phase lag). The coordinates of the joint positions defined eight body segments: the feet, lower legs, upper legs, pelvis, trunk/head, upper arms, forearms and hands. Anthropometric data (body mass, length of segments) were measured. The segment mass and the position of segmental CoM, except for the trunk, and segment moment of inertia were calculated according to Plagenhoef et al. (1983) and Looze et al. (1992). The coordinates of Th1 and the L5-S1 joint were used to determine the position of the trunk CoM during the movement according to an optimization procedure, which improved the estimated trajectory of the body CoM (Kingma et al. 1995). The mass and CoM location of the feet were recalculated to include the mass and CoM location of the shoes. The length of the base of support was inferred from the markers on the fifth metatarsophalangeal and ankle joint. The sagittal plane location of the box CoM was inferred from the three box markers. The body CoM location was calculated from the segments' mass and CoM location. The angle of each segment was calculated relative to the horizontal. Numerical differentiation (Lanczos 5-point differentiation filter) of the time histories of positions and angles yielded (angular) velocities.

Box forces

A special box, equipped with ten strain gauge force transducers was used in this study. The box consists of a solid metal frame to which two handles are mounted and to which additional weights can be attached. A (replaceable) synthetic cover (width {left-right} 0.400 m, height 0.200 m, depth {anterior-posterior} 0.265 m) is firmly attached to the frame. The handles are positioned on each side of the box, 0.131 m above the bottom and 0.046 m away from the side. The weight of the empty box (frame, handles and cover) is 6.7 kg. On each side of the frame, close to the handles, three transducers recorded forces in three orthogonal directions. Of these, the vertical force, defined as the *load force* according to Johansson and Westling (1984), and the anterior-posterior force, i.e. the *horizontal box force*, were further analysed. In each handle, two additional transducers, one at the front and one at the back of the handle, recorded the force exerted downward on the handle; the sum of those forces is the *grip force*. The analog force signals were amplified, low-pass filtered (30 Hz, 4th order), sampled (600 Hz, 12 bits) and stored in synchrony with the recorded movements by the VICON system. Off-line, the forces were digitally low-pass filtered (zero phase lag, 60 Hz, 2nd order Butterworth) and corresponding forces measured on each handle were added, yielding one net load force, one net horizontal box force and one net grip force. The *box reaction force* on the body was determined from the net load force and net horizontal box force.

Ground reaction force

A strain gauge force platform (1.0 x 1.0 m) recorded the ground reaction force in three orthogonal directions. Of these, the vertical and anterior-posterior forces, i.e. the *vertical and horizontal ground reaction force* respectively, were further analysed. The analog force signals were amplified, low-pass filtered (30 Hz, 4th order), sampled (600 Hz, 12 bits) and stored in synchrony with the recorded movements and the box force signals by the VICON system. From the distribution of the force components, the point of application (the centre of pressure CoP) of the force vector was calculated (with a maximal error of 3 mm). The ground reaction force and CoP were digitally low-pass filtered (zero phase lag, 60 Hz, 2nd order Butterworth).

Biomechanical analysis

The subjects' movements were quantified by the horizontal and vertical momentum of the CoM and by the angular momentum of the whole body. Considering the whole body as a single free body, only external forces and moments of external forces can change the linear and angular body momenta. External forces are the force of gravity, the ground reaction force ($\boldsymbol{F}_g$) and the box reaction force ($\boldsymbol{F}_b$) (Figure 7.2, left panel) and external moments are the moment of the ground reaction force about the body CoM ($\boldsymbol{M}_g$) and the moment of the box reaction force about the body CoM ($\boldsymbol{M}_b$) (Figure 7.2, right panel). To generate the upward lifting motion and counteract the decelerating effects of $\boldsymbol{F}_b$ and $\boldsymbol{M}_b$, the subjects were expected to generate additional torques at joints forming the relevant kinematic chain, such that $\boldsymbol{F}_g$ and $\boldsymbol{M}_g$ would increase prior to or simultaneously with the decrease in $\boldsymbol{F}_b$ and $\boldsymbol{M}_b$. It was assumed that the subjects actively controlled $\boldsymbol{F}_g$, through an adequate combination of joint torques, and thus accomplished the required postural adjustments in linear and angular momenta. This assumption stems from the finding that control of the magnitude and direction of $\boldsymbol{F}_g$ is, next to control of joint angular changes, crucial in the execution of contact force control tasks, tasks in which a force is exerted on the environment (Ingen Schenau et al. 1992; Jacobs and Ingen Schenau 1992a). As regulation of the contact force direction was found to rely on the generation of a particular *combination* of net torques at the joints involved (Ingen Schenau et al. 1992), it was suggested that the CNS does not control the *individual* joint torques, but rather the contact force (Jacobs and Macpherson 1996; Macpherson 1988a; 1988b).

The above mentioned biomechanical parameters were calculated as follows:

- The instantaneous horizontal and vertical body CoM momenta were calculated from the sum of respectively the horizontal and vertical momenta of all segmental CoMs.
- The instantaneous angular momentum of the whole body was calculated from the sum of the segmental angular momenta according to Toussaint et al. (1995).
- The external moment of $\boldsymbol{F}_g$ about the body CoM ($\boldsymbol{M}_g$) was calculated from $\mathbf{a}_g$ x $\boldsymbol{F}_g$ (cross-product), with $\mathbf{a}_g$ the vector from body CoM to CoP (Figure 7.2, right panel).
- The external moment of $\boldsymbol{F}_b$ about the body CoM ($\boldsymbol{M}_b$) was calculated from $\mathbf{a}_b$ x $\boldsymbol{F}_b$ (cross-product), with $\mathbf{a}_b$ the vector from the body CoM to the hand CoM (Figure 7.2, right panel).
- The horizontal positions of CoM and CoP were expressed as a percentage of the base of support length: CoM_{rel} and CoP_{rel}, with 0% at the heel and 100% at the toes.

The caudo-cranial and dorso-ventral directions were defined positive for translations (Figure 7.2, left panel) and the counter-clockwise direction was defined positive for the angular momentum and external moments of force (Figure 7.2, right panel).

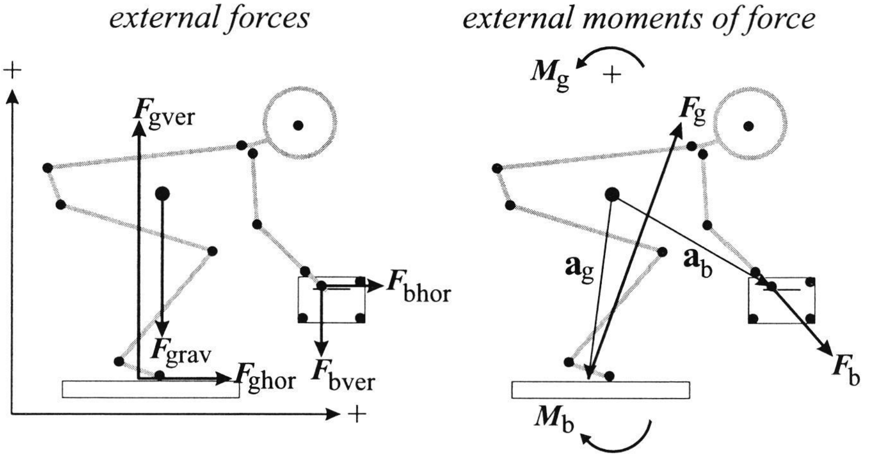


Figure 7.2
Schematic drawing of the external forces and external moments of force.
The subject (provided with eleven light-reflecting markers, the small dots), the box (provided with three markers) and the force platform are shown in the left panel. The centre of mass (CoM) of the subject is represented by the bigger dot. Three external forces are applied on the subject: the force of gravity ($\boldsymbol{F}_{grav}$), the ground reaction force ($\boldsymbol{F}_g$, split into its horizontal and vertical component: $\boldsymbol{F}_{ghor}$ and $\boldsymbol{F}_{gver}$) and the box reaction force ($\boldsymbol{F}_b$, split into its horizontal and vertical component: $\boldsymbol{F}_{bhor}$ and $\boldsymbol{F}_{bver}$). The right panel shows the external moment of $\boldsymbol{F}_g$ and $\boldsymbol{F}_b$ about the CoM ($\boldsymbol{M}_g$ and $\boldsymbol{M}_b$ respectively). $\boldsymbol{F}_{grav}$ does not exert a moment about the CoM. The vectors from the body CoM to the point of application of $\boldsymbol{F}_g$ and $\boldsymbol{F}_b$ on the body are indicated by, respectively, $\mathbf{a}_g$ and $\mathbf{a}_b$.

Definition of phases

Kinematics, box and ground reaction forces were recorded during two movement cycles. One movement cycle comprises a down- and subsequent upward phase. Figure 7.1 illustrates the method we applied to define the onset and end of each down- and upward phase. Six instants (m1 to m6), which coincided with the extremes in vertical CoM position, were marked in the time trace of the vertical body CoM momentum (right panel). The three criteria used to mark these instants were determined by visual inspection of several trials of each subject. The highest CoM position, defining the onset of the first and second downward phase, is marked by m1 and m4 (criterium: first sample at which the vertical momentum was lower than -2.5 kg•m•s^{-1}). The lowest CoM position around the end of a downward/onset of an upward phase is marked by m2 and m5 (criterium: first sample at which the vertical momentum exceeded +0.5 kg•m•s^{-1}). The end of down II/onset of up II was set at the onset of the upward load force on the box. The end of down I/onset of up I was obtained by subtracting from m2, the time difference between the onset of the upward load force and m5. The highest CoM position, defining the end of the first and second upward phase, is marked by m3 and m6 (criterium: first sample at which the vertical momentum was lower than +1.0 kg•m•s^{-1}).

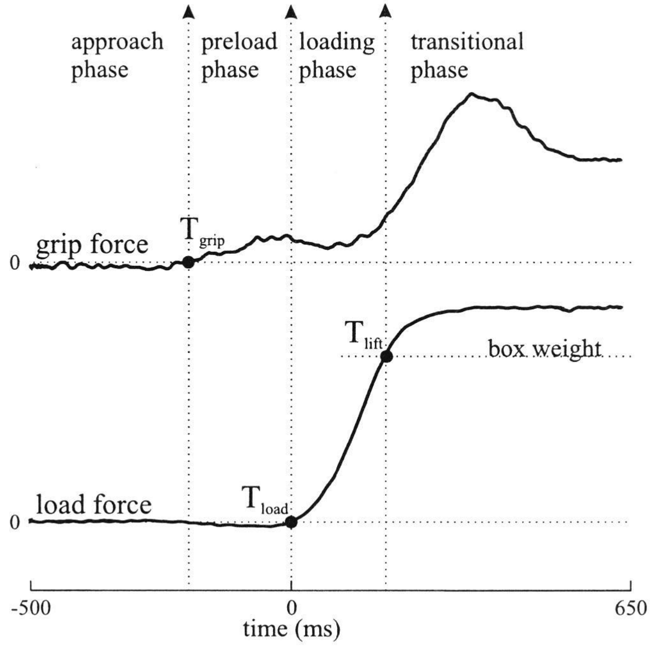


Figure 7.3
Definition of relevant phases in the loaded movement cycle, in which the box is grasped and lifted. The time traces of the load force and grip force exerted on the box (typical example) are used to define the approach, preload, loading and transitional phase. The symbols T_{grip}, T_{load} and T_{lift} mark the onset of, respectively, the preload, loading and transitional phase. The text describes the criteria used to mark the onsets.

In the loaded movement cycle in which the box was grasped and lifted, several phases were defined according to Johansson and Westling (1984). The *approach phase* is the time period from the onset of the second downward phase to the onset of the first increase in grip force (T_{grip}, Figure 7.3). The *preload phase* is the interval between T_{grip} and the onset of the first increase in load force (T_{load}). The *loading phase* is the time period between T_{load} and the instant at which the load force first exceeds the box weight (T_{lift}). This last event coincides with the onset of vertical motion of the box, i.e. lift-off. The *transitional phase* is the interval between T_{lift} and the moment at which the box reaches the intended position at chest height. The onsets of the first increase in grip force (T_{grip}) and the first increase in load force (T_{load}) were determined by visual inspection of the respective time traces in an interactive data analysis software application (Matlab^R for Windows, Version 4.0, The MathWorks Inc.). Instant T_{lift} was set automatically at the sample the load force first exceeded the box weight. The occurrence in time of various box and body related events (defined in the results section) was manually determined in Matlab by visual inspection of the respective time traces.

Data analysis and statistics

To compare the biomechanical parameters of the experimental conditions (the unloaded cycle, and the loaded cycle of the 6.7 kg-, 16.7 kg- and 6.7-*expecting 16.7*-kg-trials), time traces of each parameter of all trials (eight, eight, four and one respectively) were averaged per subject. Next, the parameter time traces of the eight subjects were averaged per condition (mean ± 1 standard error of the mean -SEM-). Each trial was synchronized in time to the end of the downward/onset of the upward phase (T_{load} in the loaded conditions). A total of 1150 ms (500 before and 650 ms after T_{load}) was analysed, according to the duration of the movement cycle that (of the 168 cycles analysed) was performed most quickly. The timing of various box and body related events and the magnitude of parameters at a defined moment were determined for each trial of each subject. Next, the values were averaged per subject and an overall value (mean ± 1 SD) was calculated for each condition. Differences in timing and magnitude of parameters between conditions were tested with a repeated measures analysis of variance or a paired-samples *t*-test. Differences were considered significant in case of $p<0.05$.

Results

I. Anticipatory postural adjustments

Characteristics of the ongoing motion and of the expected perturbation

The kinematics and kinetics of the loaded movement cycle of one subject, lifting the 6.7 kg-box, are presented in Figure 7.4. The subject bent the upper body forward and moved the whole body downward to reach for the box in the approach phase (before T_{grip}).

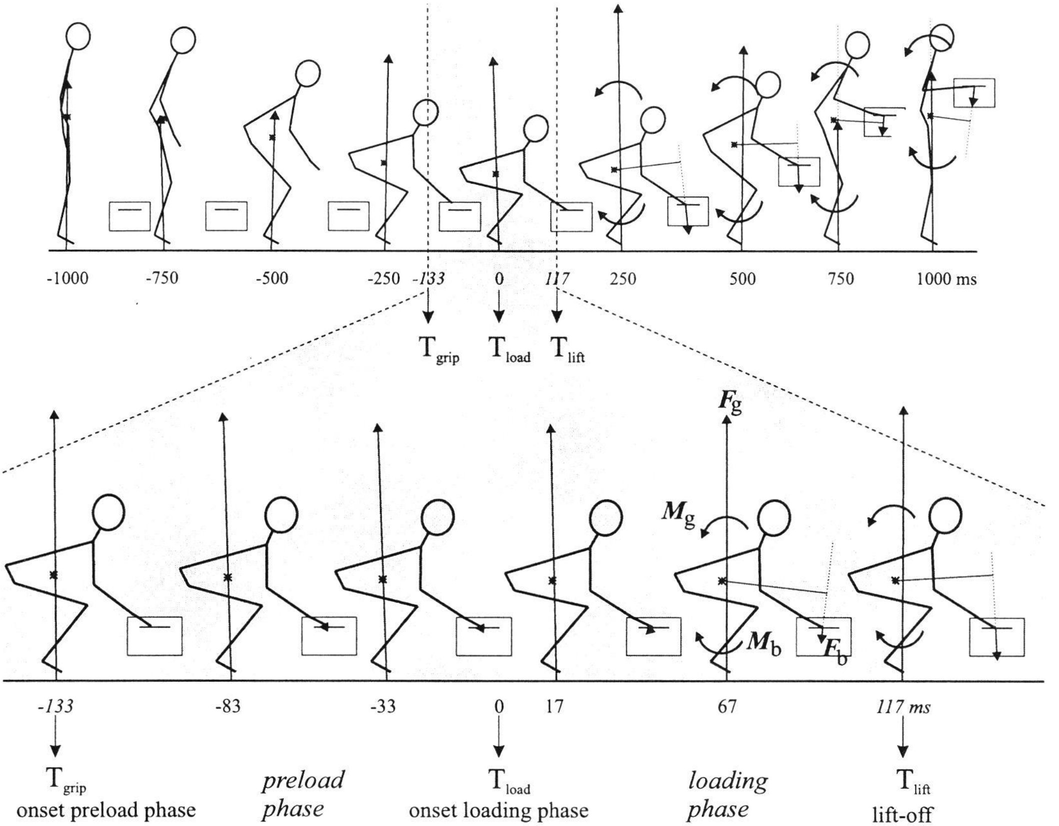


Figure 7.4
A sequence of stick-figures illustrates the characteristics of the bimanual, whole-body lifting task. One 6.7 kg-trial of one subject is presented as a sample trial. T_{grip} marks the onset of the preload phase, T_{load} the onset of the loading phase and T_{lift} the onset of the transitional phase, at box lift-off. The upper row shows the complete loaded movement cycle and the lower row zooms to the preload and loading phase. The body centre of mass (CoM) is marked by an asterisk, the upward pointing vector represents the ground reaction force ($\boldsymbol{F}_g$) and the downward pointing vector the box reaction force ($\boldsymbol{F}_b$). The direction of the moments of $\boldsymbol{F}_g$ and $\boldsymbol{F}_b$ about the CoM ($\boldsymbol{M}_g$ and $\boldsymbol{M}_b$ respectively) is indicated by curved arrows. The momentarm of the force vector is indicated by a thin line between the line of action of the force vector and the CoM.

Next, during the preload and loading phase (magnified in the lower part), only minor changes were observed in the subject's kinematics, but large ones in the box reaction force ($\boldsymbol{F}_b$). During the preload phase, the subject grasped the box handles and a small backward directed $\boldsymbol{F}_b$ occurred (i.e. the subject pushed forward on the box). During the loading phase, a downward directed $\boldsymbol{F}_b$ arose quickly, indicating that the subject applied an upward force on the box to achieve its lift-off. This downward $\boldsymbol{F}_b$ may decrease the upward body CoM momentum, required to raise the body. The onset of this perturbing $\boldsymbol{F}_{bver}$ was, by definition, at t=0 ms (T_{load}). Furthermore, the backward directed $\boldsymbol{F}_b$ attained a forward direction, thus threatening to decrease the current backward body CoM momentum. The onset of this perturbing $\boldsymbol{F}_{bhor}$ occurred 75 (±35) ms and 69 (±41) ms after T_{load} for, respectively, the 6.7 and 16.7 kg-trials. And finally, $\boldsymbol{F}_b$ exerted a clockwise directed moment about the CoM ($\boldsymbol{M}_b$), thus threatening to decrease the counter-clockwise whole-body angular momentum, required to extend the body. The onset of this perturbing moment was also, by definition, at t=0 ms. Simultaneously, the vertical component of the ground reaction force ($\boldsymbol{F}_g$), being larger than the subject's weight (Figure 7.7, upper left panel), increased the upward momentum and $\boldsymbol{F}_g$ exerted a counter-clockwise directed moment about the CoM ($\boldsymbol{M}_g$), increasing the counter-clockwise angular momentum. After box lift-off, the subject extended the upper body backward again and moved the body and box upward. $\boldsymbol{F}_b$ slowly decreased to a stable level, while $\boldsymbol{F}_g$ fluctuated notably in magnitude. The direction of both $\boldsymbol{M}_b$ and $\boldsymbol{M}_g$ did not change during this transitional phase.

Characteristics of the anticipatory postural adjustments

The main goal of the present study was to identify the leading principle(s) in the organization of postural adjustments in a bimanual, whole-body lifting task. If regulation of the horizontal CoM position is the main cue in the anticipatory postural adjustments, it can be expected that the CoM_{rel} will be displaced backward prior to the onset of the (forward directed) disturbance, and that this displacement will be more pronounced when the disturbance magnitude is larger. Figure 7.5 shows that the CoM_{rel} was not displaced backward prior to load pick-up (at t=0) in the loaded cycles (6.7 and 16.7 kg), compared with the unloaded one. Rather, the CoM_{rel} tended to be displaced forward and this shift seemed larger when the 16.7 kg-box was to be lifted. Testing average positions at t=-500 and t=-250 showed that the CoM_{rel} was positioned significantly closer to the toes in the 16.7 kg-trials, compared with the unloaded ones (-500: 58.7 ± 4.2 v. 55.2 ± 5.1%, t=3.61, p=0.009 and -250: 60.8 ± 4.2 v. 57.1 ± 5.9%, t=3.43, p=0.011). Differences in position between the 6.7 kg- v. the unloaded trials and the 6.7 kg- v. the 16.7 kg-trials were not significant at -500 and -250 ms. Despite the different time traces in the approach phase, the CoM_{rel} was at the same position in all conditions at load pick-up (58.5 ± 6.2 (unloaded), 58.3 ± 6.3 (6.7 kg) and 57.3 ± 4.3% (16.7 kg), $F_{(2,21)}$=0.60, p=0.564).

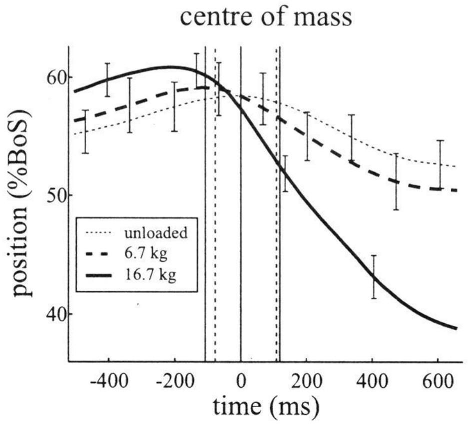


Figure 7.5
Time traces of the unloaded (thin dashed), loaded 6.7 kg (thick dashed) and loaded 16.7 kg (thick solid) movement cycle of a bimanual, whole-body lifting task for the horizontal position of the centre of mass relative to the base of support (BoS). Mean time traces ± 1 SEM (n=8) are presented. The solid vertical line at t=0 ms marks the onset of the loading phase in the loaded cycles and the corresponding instant in time in the unloaded cycle. The vertical lines at the left and right of t=0 mark, respectively, the onset of the preload phase and box lift-off in the loaded cycles of the 6.7 kg- (dashed) and 16.7 kg-trials (solid).

Pai and Patton (1997) have recently proposed that it is the combination of CoM position and velocity, rather than CoM position alone, that defines which movements are feasible in stance without loss of equilibrium. They applied an inverted pendulum model with a foot segment and an optimization algorithm to determine the set of 'safe' CoM velocity-position combinations (Pai and Patton 1997). Thus, for the present lifting task we ought to examine the combination of adjustments in horizontal CoM position ànd momentum.

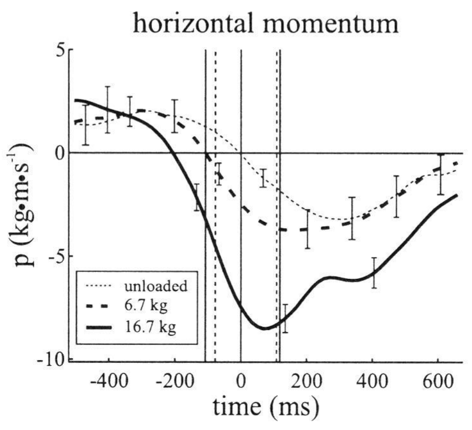


Figure 7.6
Time traces of the unloaded (thin dashed), loaded 6.7 kg (thick dashed) and loaded 16.7 kg (thick solid) movement cycle of a bimanual, whole-body lifting task for the horizontal body centre of mass momentum. Illustrational conventions as in Figure 7.5.

Figure 7.6 reveals a distinct anticipatory adjustment in the horizontal momentum. During the last part of the downward phase, this parameter showed a rapid decrease to negative values. The decrease was more pronounced for the 16.7 kg-trials. As a result, a progressively larger backward momentum was present at T_{load} in the condition in which the to be lifted load was larger (-0.1 ± 1.2 (unloaded), -2.5 ± 1.5 (6.7 kg) and -7.5 ± 2.3 kg•m•s^{-1} (16.7 kg),

$F_{(2,21)}$=43.11, p<0.001, see also the lower panel of Figure 7.8). Thus, the horizontal CoM *momentum*, rather than the *position* displayed the expected anticipatory adjustment, which seemed to be related to the future disturbance. It may be questioned, though, why this adjustment occurred that long before the onset of the disturbance.

As outlined in the introduction section and illustrated in Figure 7.4, the expected equilibrium perturbation did not only affect the horizontal CoM position and momentum. The vertical and angular momentum were also influenced. Furthermore, given the pronounced rotations of body segments in space and relative to each other (see Figure 7.4), it seems not justified to model the multi-joint lifting motion as an inverted pendulum rotating around the ankle joint. We propose to extend the analysis of the anticipatory adjustments to the vertical and angular momenta and to quantify the perturbation and relate it to the anticipatory adjustments. Then, we are able to analyse how the perturbing $\boldsymbol{F}_b$ and $\boldsymbol{M}_b$ affected the linear CoM momenta and whole-body angular momentum during the lifting task and in what way subjects counteracted this perturbation by generating anticipatory adjustments in $\boldsymbol{F}_g$ and $\boldsymbol{M}_g$.

Figure 7.7
Time traces of the unloaded (thin dashed), loaded 6.7 kg (thick dashed) and loaded 16.7 kg (thick solid) movement cycle of a bimanual, whole-body lifting task for the vertical ground reaction force ($\boldsymbol{F}_g$) minus the force of gravity (upper left panel), the vertical body centre of mass (CoM) momentum (upper right panel), the moment of $\boldsymbol{F}_g$ about the CoM ($\boldsymbol{M}_g$, middle left panel), the whole-body angular momentum (middle right panel), the horizontal $\boldsymbol{F}_g$ (lower left panel) and the horizontal CoM momentum (lower right panel). In addition, the left panels show the time traces of the vertical box reaction force ($\boldsymbol{F}_b$), the moment of $\boldsymbol{F}_b$ about the CoM ($\boldsymbol{M}_b$) and the horizontal $\boldsymbol{F}_b$ for the 6.7 kg (thick dashed) and 16.7 kg-trials (thick solid). Illustrational conventions as in Figure 7.5.

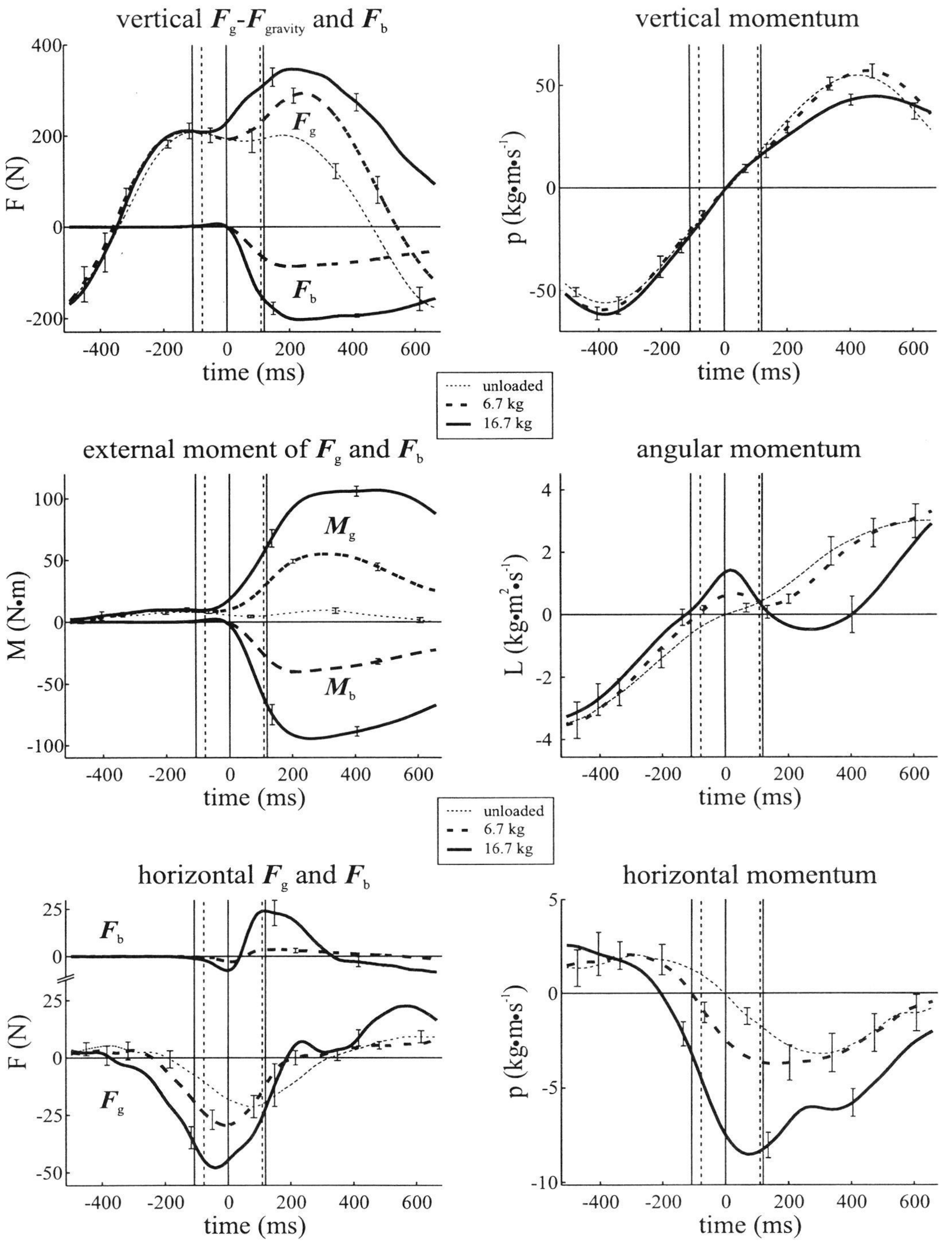
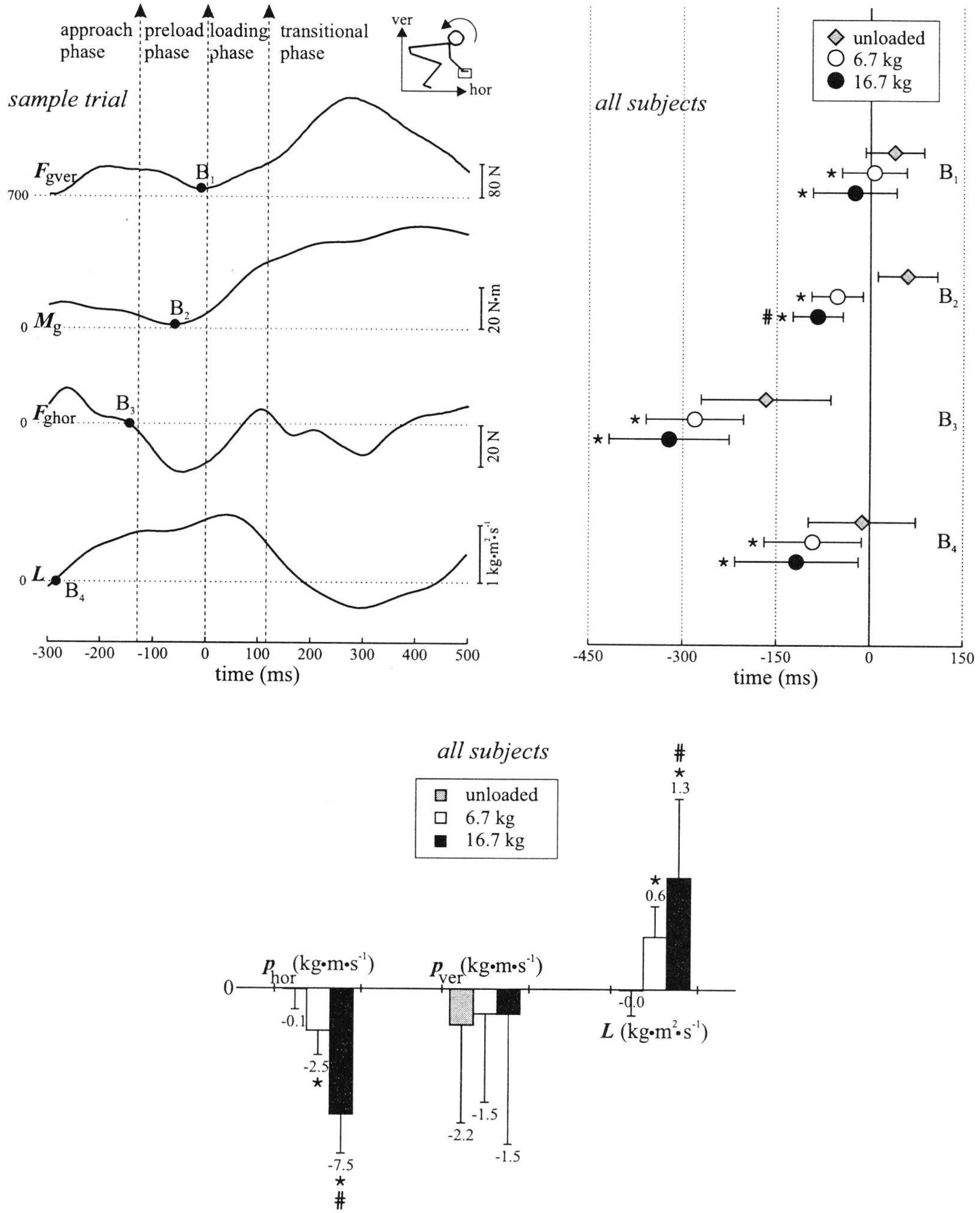

vertical F_g-$F_{gravity}$ and F_b
F (N)
time (ms)
F_g
F_b
vertical momentum
p (kg•m•s^{-1})
time (ms)
unloaded
6.7 kg
16.7 kg
external moment of F_g and F_b
M (N•m)
time (ms)
M_g
M_b
angular momentum
L (kg•m^2•s^{-1})
time (ms)
unloaded
6.7 kg
16.7 kg
horizontal F_g and F_b
F (N)
time (ms)
F_b
F_g
horizontal momentum
p (kg•m•s^{-1})
time (ms)

Figure 7.7 displays the onset, direction and magnitude of the perturbing $\boldsymbol{F}_b$ and $\boldsymbol{M}_b$. The onset of the downward $\boldsymbol{F}_{bver}$ and clockwise $\boldsymbol{M}_b$ occurred at T_{load} and the onset of the forward $\boldsymbol{F}_{bhor}$ around +70 ms (left panels). To perform the multi-joint lifting motion and counteract the decelerating effects of $\boldsymbol{F}_b$ and $\boldsymbol{M}_b$, it can be expected that subjects will increase $\boldsymbol{F}_{gver}$ and $\boldsymbol{M}_g$ just prior to or simultaneously with the decrease in $\boldsymbol{F}_{bver}$ and $\boldsymbol{M}_b$, and decrease $\boldsymbol{F}_{ghor}$ just prior to or simultaneously with the increase in $\boldsymbol{F}_{bhor}$. Figure 7.7 demonstrates that $\boldsymbol{F}_{gver}$ and $\boldsymbol{M}_g$ indeed increased around T_{load}, but that $\boldsymbol{F}_{ghor}$ did not display the expected decrease (to more negative values) around the onset of the forward $\boldsymbol{F}_{bhor}$ (left panels). On the contrary, the backward $\boldsymbol{F}_{ghor}$ was increasing again to less negative values at that instant. The onsets of the increase in $\boldsymbol{F}_{gver}$ (B_1) and $\boldsymbol{M}_g$ (B_2) relative to the onset of the loading phase are shown in the upper panels of Figure 7.8. The anticipatory nature of both events is illustrated in the upper right panel. In both loaded conditions, the onsets occurred significantly earlier compared with the unloaded one. Moreover, the onset of the increase in $\boldsymbol{M}_g$ was significantly advanced in the 16.7 kg-trials, as opposed to the 6.7 kg-trials. The onset of a backward $\boldsymbol{F}_{ghor}$ (B_3) preceded the onset of the forward $\boldsymbol{F}_{bhor}$ with 356 (±63) ms (6.7 kg) and 390 (±79) ms (16.7 kg). Thus, the anticipatory adjustments in $\boldsymbol{F}_{gver}$ and $\boldsymbol{M}_g$, but not $\boldsymbol{F}_{ghor}$, seemed to be time-locked to the perturbation onset.

Figure 7.8
The upper left panel presents time traces of the vertical ground reaction force ($\boldsymbol{F}_{gver}$), the moment of the ground reaction force about the centre of mass ($\boldsymbol{M}_g$), the horizontal ground reaction force ($\boldsymbol{F}_{ghor}$) and the whole-body angular momentum ($\boldsymbol{L}$) for the sample trial (6.7 kg-box). The vertical dotted lines indicate, from left to right, the onset of the preload, loading and transitional phase. The onsets of the increase in $\boldsymbol{F}_{gver}$ (B_1), the increase in $\boldsymbol{M}_g$ (B_2), a backward $\boldsymbol{F}_{ghor}$ (B_3) and a counter-clockwise directed $\boldsymbol{L}$ (B_4) are marked. The mean timing ± 1 SD (n=8) of these events with respect to the onset of the loading phase (T_{load}) and with respect to the corresponding instant in the unloaded cycle (t=0) is displayed in the upper right panel. The mean magnitude ± 1 SD (n=8) at T_{load} of the horizontal centre of mass momentum ($\boldsymbol{p}_{hor}$), the vertical centre of mass momentum ($\boldsymbol{p}_{ver}$) and the whole-body angular momentum ($\boldsymbol{L}$) is shown in the lower panel. Significant differences in timing and magnitude are marked: * for the 6.7 and 16.7 kg-condition v. the unloaded cycle and # for the 6.7 v. 16.7 kg-trials.


approach phase
preload phase
loading phase
transitional phase
ver
hor
sample trial
all subjects
unloaded
6.7 kg
16.7 kg
time (ms)
all subjects
unloaded
6.7 kg
16.7 kg
0.1
-2.5
-7.5
-2.2
-1.5
-1.5
-0.0
0.6
1.3

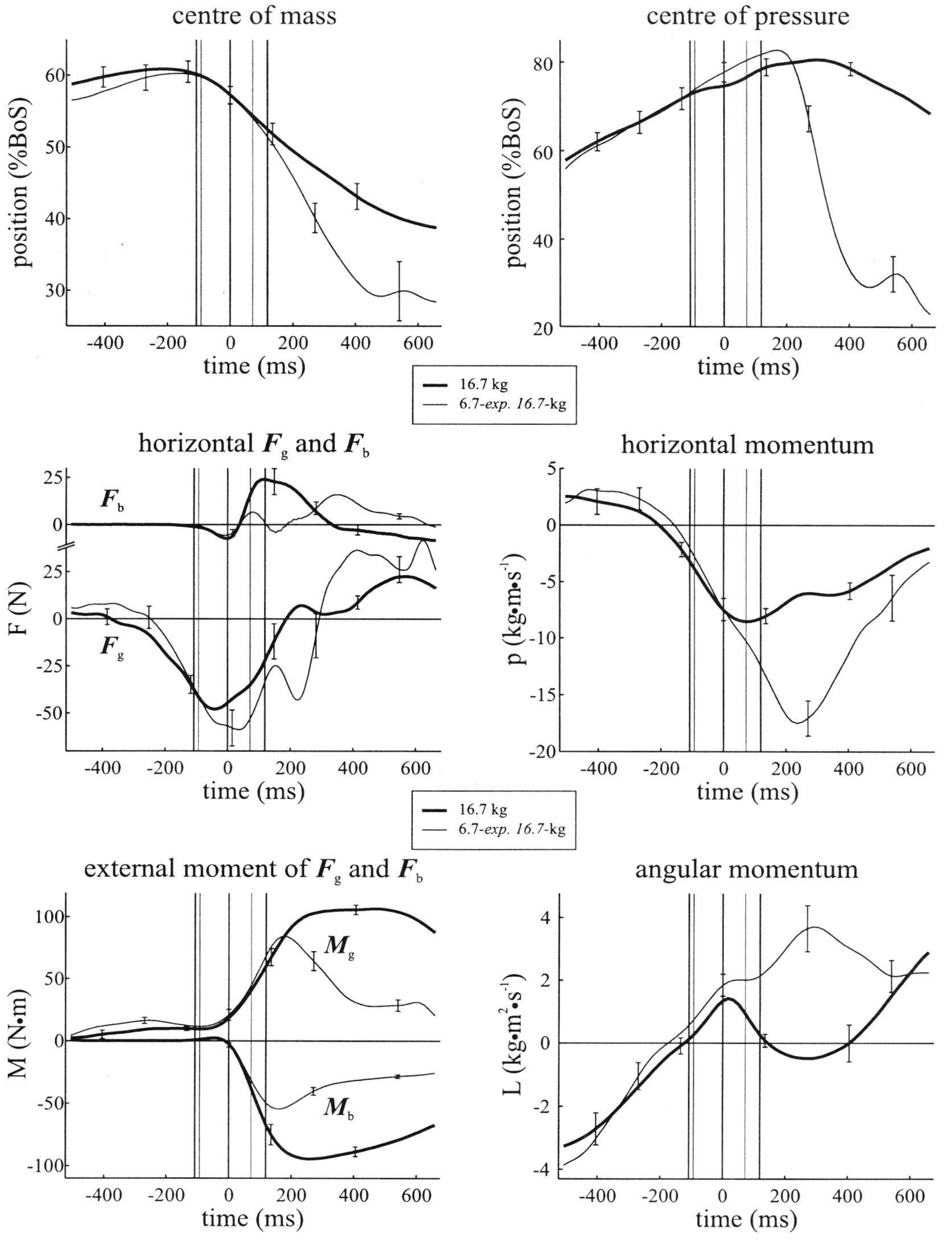
The effect of the anticipatory adjustments in $\boldsymbol{F}_g$ and $\boldsymbol{M}_g$ on the linear and angular momenta is illustrated in the right panels of Figure 7.7 and the lower panel of Figure 7.8. Because of the early onset of the adjustment in $\boldsymbol{F}_{ghor}$, the backward horizontal momentum was already at the onset of the loading phase significantly larger in both loaded conditions than in the unloaded one and significantly larger in the 16.7 kg- than in the 6.7 kg-trials (Figure 7.8, lower panel). The magnitude of the vertical momentum was not significantly different between conditions at T_{load}. Despite the anticipatory increase in $\boldsymbol{M}_g$, the angular momentum showed a substantial dip after load pick-up for both loaded conditions (Figure 7.7, middle right panel). Apparently, $\boldsymbol{M}_g$ could not increase fast enough to neutralize the quickly decreasing $\boldsymbol{M}_b$. The dip in angular momentum appeared to be expected, though, because the onset of a counter-clockwise angular momentum (B_4) occurred significantly earlier in both loaded cycles (Figure 7.8, upper panels). Thus, at T_{load}, the magnitude of the counter-clockwise angular momentum was progressively larger with increasing perturbation (i.e. clockwise $\boldsymbol{M}_b$) magnitude (Figure 7.8, lower panel).

II. Compensatory postural responses

The trials in which the 6.7 kg-box was lifted, while the 16.7 kg-box was expected were characterized by a disturbance of a smooth ongoing motion in many cases. Task performance pointed to an overestimation of the to be lifted load, since subjects initially moved up- and backward in a brusque motion. All subjects managed to lift the box to the defined end position, although this took a little more time than in the undisturbed trials for most of them.

Figure 7.9

Time traces of the loaded movement cycle of the 16.7 kg (thick) and 6.7-*expecting 16.7*-kg-trials (thin) in a bimanual, whole-body lifting task for the horizontal position of the centre of mass (CoM) relative to the base of support (BoS) (upper left panel), the horizontal position of the centre of pressure relative to the BoS (upper right panel), the horizontal ground reaction force ($\boldsymbol{F}_g$) (middle left panel), the horizontal CoM momentum (middle right panel), the moment of $\boldsymbol{F}_g$ about the CoM ($\boldsymbol{M}_g$, lower left panel) and the whole-body angular momentum (lower right panel). In addition, the middle and lower left panels show the time traces of, respectively, the horizontal box reaction force ($\boldsymbol{F}_b$) and the moment of $\boldsymbol{F}_b$ about the CoM ($\boldsymbol{M}_b$) for the 16.7 kg (thick) and 6.7-*expecting 16.7*-kg-trials (thin). Mean time traces ± 1 SEM (n=8) are presented. The solid vertical line at t=0 ms marks the onset of the loading phase. The vertical lines at the left and right of t=0 mark, respectively, the onset of the preload phase and box lift-off in the 16.7 kg- (thick) and 6.7-*expecting 16.7*-kg-trials (thin).


centre of mass
position (%BoS)
time (ms)
centre of pressure
position (%BoS)
time (ms)
16.7 kg
6.7-*exp. 16.7*-kg
horizontal F_g and F_b
F_b
F_g
F (N)
time (ms)
horizontal momentum
p (kg•m•s^{-1})
time (ms)
16.7 kg
6.7-*exp. 16.7*-kg
external moment of F_g and F_b
M_g
M_b
M (N•m)
time (ms)
angular momentum
L (kg•m^2•s^{-1})
time (ms)

Two subjects experienced such a large equilibrium disturbance, that they had to make a compensatory step backward (both with the left foot) to prevent falling. Three subjects showed major disturbances which they compensated without making a compensatory step and the other three subjects completed the task without gross signs of imbalance.

Characteristics of the compensatory postural responses

Analogous to the questions posed with respect to the anticipatory postural adjustments, it can be questioned whether regulation of the horizontal CoM position is the main cue in the organization of the compensatory responses. If this would be the case, it can be expected that the overshoot in backward CoM momentum (due to overestimating the load to be lifted) will be reduced instantly, such that the backward shifting CoM_{rel} will not exceed the rear margin of the base of support. To achieve this, the backward $\boldsymbol{F}_{ghor}$ that is normally present after lift-off (see Figure 7.7, lower left panel) should be immediately reversed to a forward direction.

Figure 7.9 shows that subjects prepared to lift the 16.7 kg-box in the trials in which, without their knowledge, the 6.7 kg-box was presented. Thus, they expected the perturbing $\boldsymbol{F}_b$ and $\boldsymbol{M}_b$ associated with that load. The actual $\boldsymbol{F}_b$ and $\boldsymbol{M}_b$ were smaller, however, and the decelerating effect of $\boldsymbol{F}_b$ and $\boldsymbol{M}_b$ on the linear and angular momenta was, thus, smaller than expected. The resulting overshoots in linear and angular momenta (middle and lower right panel) were compensated by quick adjustments in $\boldsymbol{F}_{ghor}$ and CoP. Surprisingly, these responses were aimed at reducing the overshoot in angular momentum in the first place and not the overshoot in horizontal momentum. The CoP displayed a pronounced backward shift, starting 111 (±18) ms after box lift-off (upper right panel) and $\boldsymbol{F}_{ghor}$ showed a sudden reversal, 94 (±31) ms after box lift-off, of the trend towards less negative values and instead attained more negative values again (middle left panel). Both events reduced the momentarm of the $\boldsymbol{F}_g$ vector with respect to the CoM, thereby limiting the much larger $\boldsymbol{M}_g$ that would have occurred had the CoP followed the path observed in the undisturbed 16.7 kg-trials (lower left panel). The positive $\boldsymbol{M}_g$ decreased to less positive values, starting 102 (±24) ms after box lift-off, such that 136 (±97) ms later the counter-clockwise $\boldsymbol{M}_g$ no longer exceeded the clockwise $\boldsymbol{M}_b$, yielding a deceleration of the large counter-clockwise angular momentum (lower right panel). The forward $\boldsymbol{F}_{ghor}$ that was required to decrease the large backward momentum was generated first 233 (±36) ms after lift-off (middle left panel). The forward $\boldsymbol{F}_{bhor}$ assisted, though, in reducing the backward CoM momentum.

Discussion

Control of the whole-body angular momentum prevails in the postural adjustments in the bimanual, whole-body lifting task

Applying a global mechanical analysis of the lifting task, we were able to examine how the expected perturbation (quantified by $\boldsymbol{F}_b$ and $\boldsymbol{M}_b$) affected the whole-body motion (in terms of linear CoM momenta and whole-body angular momentum) during task execution and in what way subjects counteracted this perturbation by generating anticipatory adjustments in $\boldsymbol{F}_g$ and $\boldsymbol{M}_g$. Also, this approach enabled us to fully analyze the mechanics of the postural reactions to an unexpected equilibrium disturbance that occurred after box lift-off. To our opinion, this approach was essential to examine the control of equilibrium in a whole-body lifting task, because the equilibrium disturbance did not just affect the horizontal position of the CoM with respect to the base of support, nor the horizontal CoM velocity (or momentum).

The global mechanical analysis of the lifting task characteristics suggested that regulating the horizontal CoM position, by $\boldsymbol{F}_{ghor}$, was not the leading cue in the organization of the postural adjustments that counteracted the (un-)expected equilibrium disturbance. Rather, the regulation of the angular momentum, by $\boldsymbol{M}_g$, appeared to take precedence. This suggestion emerges from two observations. Firstly, the onset of the anticipatory change in $\boldsymbol{F}_{ghor}$ preceded the onset of the perturbing $\boldsymbol{F}_{bhor}$ with more than 350 ms, while the two other parameters that regulate the multi-joint lifting motion, $\boldsymbol{F}_{gver}$ and $\boldsymbol{M}_g$, did show the expected anticipatory change close to the onset of the perturbing $\boldsymbol{F}_{bver}$ and $\boldsymbol{M}_b$. Such an early anticipation in $\boldsymbol{F}_{ghor}$ seems inefficient, for anticipatory actions are essentially a source of perturbation too and should thus occur only shortly before or simultaneously with the perturbing forces (Oddsson and Thorstensson 1986). It may, therefore, be questioned whether the anticipatory changes in $\boldsymbol{F}_{ghor}$ were actually directed at counteracting the perturbing $\boldsymbol{F}_{bhor}$ or whether these adjustments served another goal. We will provide evidence in favour of the latter in the next paragraph. The second observation was that the compensatory postural responses were initially directed at reducing the overshoot in angular momentum and not the overshoot in horizontal momentum.

Mechanical constraints influence the organization of postural adjustments in the lifting task

The first question that ought to be addressed in this discussion is why control of the angular momentum took precedence over regulation of the horizontal CoM position and momentum in the postural adjustments in this lifting task. Were the anticipatory adjustments in $\boldsymbol{F}_{\text{ghor}}$ aimed at counteracting the perturbing $\boldsymbol{F}_{\text{bhor}}$ or did they perhaps serve the anticipatory changes in angular momentum? Why did the compensatory postural responses not reduce the overshoots in backward CoM momentum and counter-clockwise angular momentum at the same time? The answers to these questions lie within the mechanical constraints associated with the multi-joint lifting movement. First we will demonstrate that the anticipatory changes in $\boldsymbol{F}_{\text{ghor}}$ were *not* aimed at counteracting the perturbing $\boldsymbol{F}_{\text{bhor}}$, but at creating a condition in which the required angular momentum could be generated. Then we will explain why the overshoots in horizontal and angular momentum could not be reduced at the same time.

To generate the upward motion of body and box and the extension of the whole body towards an upright posture, the subjects ought to increase $\boldsymbol{F}_{\text{gver}}$ and $\boldsymbol{M}_{\text{g}}$ around the onset of the loading phase, i.e. close to the onset of the decrease in $\boldsymbol{F}_{\text{bver}}$ and $\boldsymbol{M}_{\text{b}}$. These events were indeed found and their onsets suggested that they were controlled in anticipatory fashion. If, simultaneously, subjects would have further decreased $\boldsymbol{F}_{\text{ghor}}$ to counteract the disturbing increase in $\boldsymbol{F}_{\text{bhor}}$, the generation of a counter-clockwise $\boldsymbol{M}_{\text{g}}$ would have been seriously disturbed. To visualize this, just rotate the slightly backward directed $\boldsymbol{F}_{\text{g}}$ vector in the lower part of Figure 7.4 to a more backward direction. The vector would then no longer run in front of the CoM and $\boldsymbol{M}_{\text{g}}$ would attain a clockwise direction, failing to increase the counter-clockwise angular momentum required at the onset of the loading phase. Thus, a 'mechanical conflict' was present between generating a large backward $\boldsymbol{F}_{\text{ghor}}$ and a counter-clockwise $\boldsymbol{M}_{\text{g}}$ (see also chapter 5; Toussaint et al. 1995). As apparent from the results, in solving the conflict priority was given to the generation of the counter-clockwise $\boldsymbol{M}_{\text{g}}$ at and after the onset of the loading phase. The development of a backward $\boldsymbol{F}_{\text{ghor}}$ long before the onset of the loading phase can be regarded as one of the measures that facilitated this generation. The backward $\boldsymbol{F}_{\text{ghor}}$ induced (with a delay) a backward CoM shift during and after the loading phase (Figure 7.5). This was combined with a forward shift of the CoP, bringing it anterior to the CoM (the role of the CoP with respect to anticipatory postural adjustments in whole-body lifting has been described in detail in chapter 5). Thus, a positive momentarm of the $\boldsymbol{F}_{\text{g}}$ vector with respect to the CoM was created, yielding a counter-clockwise $\boldsymbol{M}_{\text{g}}$. All three events (backward F_{ghor}, backward CoM shift and forward CoP shift) were related to the required magnitude of M_g, facilitating a progressively stronger increase in M_g with increasing load. Thus, it was not possible to generate an anticipatory postural adjustment in F_{ghor} to counteract the perturbing F_{bhor}, because this was not compatible with the generation of the counter-clockwise M_g required to increase the counter-clockwise angular momentum. We propose to regard the early onset of a backward F_{ghor} as a mechanical event that facilitated this increase, rather than an advanced anticipatory postural adjustment. The early onset of and increase in backward CoM momentum were the result of the early onset of the backward F_{ghor} and should thus neither be considered an advanced anticipatory postural adjustment.

A similar mechanical conflict occurred with respect to the compensatory adjustments. The initial reactions could not reduce the overshoot in both horizontal and angular momentum, because the generation of the required forward F_{ghor} would be at variance with the necessary limitation of the counter-clockwise M_g (opposite to the conflict described above). The results section has shown that in solving this conflict priority was given to the limitation of the counter-clockwise M_g (and thereby the counter-clockwise angular momentum) and that the generation of a forward F_{ghor} was delayed. The precedence of angular over horizontal momentum regulation is further supported by the observation that two subjects made a step backward to prevent falling. Compensatory stepping can be considered as a temporary interruption of the relation between CoM position and base of support in order to achieve the limitation in counter-clockwise M_g. For stepping backward quickly brought the CoP_{rel} closer to the backward moving CoM_{rel} (see both time traces in Figure 7.9). This effectively reduced the momentarm of the F_g vector with respect to the CoM and, thus, the counter-clockwise M_g. The corrective reactions after an unexpected equilibrium disturbance during whole-body lifting have been described in detail by Toussaint et al. (submitted). The authors demonstrated that for the eight subjects who stepped in response to the perturbation, the sharp rearward CoP shift (starting 177 ± 28 ms after box lift-off) was crucial in reducing the overshoot in counter-clockwise angular momentum and, thus, in regaining equilibrium. In addition, they found that the overshoot in backward CoM momentum was first corrected after completion of the step at the time the upward motion was almost completed (about 600 ms after lift-off) (Toussaint et al. submitted).

It should be noted that the CoP played an important part in solving the mechanical conflict with respect to both the anticipatory and compensatory postural adjustments. In doing so, the CoP_{rel} deliberately deviated from the CoM_{rel} to increase or limit $\boldsymbol{M}_g$. Thus, in whole-body movements with pronounced segmental rotations (i.e. considerable fluctuations in whole-body angular momentum), the CoP_{rel} can by definition not completely follow the CoM_{rel}. As pointed out in other studies (chapters 5; Toussaint et al. 1995; Toussaint et al. submitted), we regard the CoP as a crucial variable in the control of movement and equilibrium in whole-body movements, a variable that is actively regulated by activity of ankle muscles (chapter 5; Crenna and Frigo 1991; Okada and Fujiwara 1984).

Control of the trunk/head angular momentum might be crucial to maintain equilibrium

The second question that ought to be addressed is why control of the whole-body angular momentum would be important in the organization of the postural adjustments in this task. In what way does this secure equilibrium? As defined in the methods section, the angular momentum of the whole body is composed of the angular momenta of the eight body segments. Given the large moment of inertia and a considerable angular velocity, the trunk/head segment is the main determinant of the whole-body angular momentum. The median correlation coefficient between the time traces of whole-body angular momentum and trunk/head angular momentum during the unloaded and subsequent loaded movement cycles was 0.984 (range 0.789 to 0.997, $n=104$). The minimum and maximum amplitude of the trunk/head angular momentum were, respectively, 74 ± 11% and 71 ± 13% of the minimum and maximum amplitude of the whole-body angular momentum ($n=104$). Thus, adjustments in whole-body angular momentum largely reflect adjustments in trunk/head angular momentum. The necessity to control the (angular) position of the trunk in order to maintain equilibrium was demonstrated before, although mostly with respect to upper-body tasks performed in stance. Gurfinkel et al. (1981) showed that the trunk position was the main parameter regulated for the maintenance of the vertical posture, when this posture was disturbed by rhythmic tilting of the supporting platform. The muscles crossing the ankle joint were found to be responsible for this regulation. The authors demonstrated that vestibular, nor visual, nor ankle muscle length information triggered the postural leg muscle activity. They suggested that proprioceptive information of the lumbar spine and/or feet might be crucial in signalling the disturbance of the trunk position (Gurfinkel et al. 1981).

Acknowledgements
The first author received a travel grant to Stockholm from the Netherlands Organization for Scientific Research (NWO). The authors wish to acknowledge the helpful comments of Gerrit Jan van Ingen Schenau, Jaap van Dieën, Idsart Kingma and Lieke Peper on earlier versions of the manuscript and the useful advice of Ingmarie Apel with respect to drawing the figures.

Chapter 8

Epilogue

Investigating the control of equilibrium in a bimanual, whole-body lifting task and other complex, multi-joint movements

The investigation of the control of equilibrium in a bimanual, whole-body lifting task adopted its theoretical approach initially from the investigation of the control of equilibrium in tasks with a static character. In these tasks, the upper body is involved in the primary, goal directed action, whereas the lower limbs only support the body and perform the necessary postural adjustments. Two themes take up an important position in the theoretical concepts of static equilibrium control: (1) the regulation of the horizontal position of the CoM is the leading principle in the organization of postural adjustments and (2) two separate controllers are operative in the execution of voluntary movements, one regulating the primary movement, the other equilibrium (Massion 1992; 1994).

Using the traditional approach to the investigation of equilibrium control, chapters 3 and 4 indeed provided results which supported the view that the regulation of the horizontal CoM position is the leading principle in the organization of associated postural adjustments. However, applying the 'new' biomechanical approach to investigate the control of equilibrium from a dynamic perspective and acknowledging the importance of segmental and whole-body rotations, chapters 5, 6 and especially 7 have demonstrated that the first theme is not simply applicable to the organization of postural adjustments in bimanual, whole-body lifting. The integrated control of the primary lifting motion and of equilibrium appeared to require more than the regulation of the horizontal CoM position. To accomplish the necessary (combination of) horizontal and angular momenta, a certain configuration of external forces with respect to the CoM was found essential. Moreover, expected and unexpected disturbances of equilibrium were found to be counteracted primarily by changes in the angular momentum and not by adjustments of the horizontal CoM position or velocity. Thus, control of the amount of rotation of the whole-body (a dynamic parameter) appeared to take precedence over control of the horizontal CoM position (a static parameter). The whole-body angular momentum was shown to be regulated by the moment exerted by the ground reaction force about the CoM. This so-called external moment was, in turn, strongly influenced by the point of application of the ground reaction force, the centre of foot pressure (CoP). Finally, the CoP was demonstrated to be directly related to the torque about the ankle joint, generated by the muscles crossing this joint. Thus, the crucial role that muscles crossing the ankle joint were found to fulfil in the control of equilibrium in static tasks (Aruin and Latash 1995a; Burleigh et al. 1994; Cordo and Nashner 1982; Crenna et al. 1987; Horak and Nashner 1986; Massion et al. 1993; Oddsson and Thorstensson 1987) can be also explained with the biomechanical approach presented in this thesis.

With regard to the second theme, Toussaint et al. (in press 1997a; in press 1997b) have recently questioned the strict dichotomy in the control of movement and equilibrium on the basis of a similar global mechanical analysis of bimanual, whole-body lifting tasks. The authors demonstrated that the forward CoM shift associated with load pick-up was partly counteracted by postural adjustments prior to grasping the object and partly thereafter, while performing the upward lifting movement. This finding was at variance with the commonly observed postural adjustment of an expected disturbance of the horizontal CoM position *prior* to the disturbance (Belen'kii et al. 1967). The 'delay' of the postural adjustments could be interpreted as serving a confinement of the net torque at the lumbo-sacral joint just after load pick-up. The generation of the backward directed ground reaction force, required to promote the backward CoM motion, appeared to be postponed in part until the onset of the upward lifting motion. Thus, the occurrence of the backward directed ground reaction force coincided with the occurrence of the peak torque at the lumbo-sacral joint, thereby limiting the magnitude of this torque (Toussaint et al. in press 1997b). The authors suggested that this was only possible if "the control of the anticipatory postural adjustments is part and parcel of the control of the lifting movement itself" (Toussaint et al. in press 1997a; in press 1997b). The results of chapter 7 support this suggestion. The proposal of Toussaint et al. was based on and in line with a study of Aruin and Latash (1995b), in which it was suggested that "postural adjustments may not be a mere addition to a 'voluntary motor command', but an inherent part of it". According to this view, the motor commands giving rise to postural adjustments and those regulating the primary, focal motion are generated by one common controller, rather than by two separate ones (Aruin and Latash 1995b). The neurophysiological substrates of this common controller are not discussed, though.

Although the present thesis questions the validity of the central themes of 'static equilibrium control' for the investigation of 'dynamic' tasks, the formulation of themes that are specific for the control of movement and equilibrium in a bimanual, whole-body lifting task in particular and complex, multi-joint movements in general, is still in its infancy. Future research could be instrumental to word such themes. The investigation of other dynamic tasks (walking, rising from a seat, manipulative tasks during locomotion) by means of the presented global mechanical analysis could contribute to the formulation of the basic neural principles and mechanical constraints in the control of movement and equilibrium in complex, multi-joint movements. A multi-disciplinary approach of research is advocated, combining biomechanical considerations, such as provided in this thesis, with neurophysiological facts.

The neuro-muscular coordination of occupational tasks in relation to the loading of the lumbar spine

In the largest part of this thesis, a bimanual, whole-body lifting task is investigated to learn about the organization of associated and anticipatory postural adjustments in a dynamic, multi-joint movement in general. However, knowledge about the neuro-muscular coordination of occupational tasks, such as bimanual load lifting, may be instrumental in the search for factors that cause or influence the high prevalence of low-back pain in the industrialized world (estimated life-time prevalence is 55% to 87%, Nicolaisen and Jorgensen 1985; Riihimäki 1985; Videman et al. 1984), for it is known that changes in the lumbar lordosis and posture directly influence not only the loading of the lumbar spine, but also the stiffness and stability of the passive tissue (Shirazi-Adl and Parnianpour 1996). Many studies have addressed the problem of work-related disorders and injuries from a biomechanical, physiological, epidemiological or psychophysical point of view (e.g. Anderson 1986; Dieën and Toussaint 1993; Looze et al. 1992b; 1994; 1995; McGill and Norman 1986; Snook and Irvine 1969). Just recently, a few groups have started to study the neuro-muscular coordination of lifting tasks (Burgess-Limerick et al. 1993; 1995; Oddsson et al. 1995; Scholz 1993a; 1993b; Scholz and McMillan 1995; Scholz et al. 1995; Toussaint et al. 1992; 1995; in press 1997a; in press 1997b; submitted). Oddsson et al. (1995), for example, found that an external perturbation to equilibrium during a lifting task elicited a conflict between the postural and focal motor commands, that simultaneously required different functions of the Erector Spinae muscle. This conflict between excitating and inhibiting motor commands may result in a hazardous redistribution of loads among active and passive components of the lumbar spine. Burgess-Limerick et al. (1993) and Scholz and colleagues (1993b; 1995) applied the relative phase of motion between joints as a collective variable to describe the inter-joint coordination during bimanual, whole-body squat lifting. The latter authors demonstrated that the relative phase of motion between the knee and lumbo-sacral joint changed as a function of load during the phase of lifting, but not during the phase of lowering the load: with increasing load, the extension of the lumbar spine occurred increasingly later in time, relative to the extension of the knees. Dieën et al. (1996) found that the inter-joint coordination in both squat and stoop lifting did not change across repetitive lifts when fatigue developed, except for the relative timing between knee and hip extension in squat lifting. The authors suggested that adaptation of peripheral control properties, such as the sensitivity of muscle spindles, might have been sufficient to accommodate the fatigue induced changes of the input-output characteristics of the muscles, in order to maintain the stereotyped kinematic pattern. The present thesis has demonstrated that the absence of load knowledge, and the ensuing overestimation of the load to be lifted, can lead to an increased mechanical load on the lumbar spine and to an increased risk of losing balance in lifting tasks, events that may both contribute to the higher risk of low-back injury in manual materials handling tasks. Future research on the neuro-muscular coordination of occupational tasks, extending the research presented in this thesis, may contribute to the further identification of factors that cause or influence the high prevalence of low-back pain in the industrialized world.

Control of posture and equilibrium in relation to motor development and learning

Knowledge about the basic neurophysiological and mechanical principles of the control of posture and equilibrium can be of value for physical and other therapists, working in the area of rehabilitation and motor learning. As pointed out in the literature reviewed (chapter 1) and the studies reported in this thesis, skilful performance of many daily tasks requires adequate control of posture and equilibrium. Moreover, postural control and its development was argued to play a very important role in the development of successful, autonomous actions in children (Reed 1989). Adequate postural control of the head-neck-trunk system, for example, is a necessary prerequisite for the development of smooth visual exploration and functional reaching (Reed 1989). The provision of external postural support to a child's body was demonstrated to allow the emergence of functional actions that did not yet belong to the repertoire of the child, but were indeed in accordance with the new postural control abilities (Gustafson 1984). In line with this study, Reed (1989) stated that it cannot be said that an infant really acts in a functional and adaptive way if it requires external support to act effectively. In other words, self-generated focal movements should be accompanied by self-generated postural actions. In the same line of thought, Hirschfeld (1996) advocated the training or re-learning of daily activities in physical therapy *within* the correct daily postural context, that is without external postural support (if possible). In doing so, the focal movements are (re-)learned in conjunction with the required postural actions. This implies that (physical) therapists should let the patient initiate and complete the motor task on his own, inviting the CNS to find the optimal solution for the motor problem (Hirschfeld 1996). In this way, feed-forward control can develop, which is a prerequisite for fast, smooth and accurate motor behaviour. It is the task of the (physical) therapist to identify goals, design the learning situation, and allow the CNS to exercise and develop the feed-forward control of movement and equilibrium (Hirschfeld 1996; Latash and Anson 1996). Future studies on the control of equilibrium in bimanual, whole-body lifting tasks or other complex, multi-joint movements could investigate to what extent the basic mechanical principles identified in able-bodied humans hold for people whose ability to perform voluntary movements is impaired for some reason. Thus, not only a deeper understanding of the control of equilibrium in non-impaired multi-joint movements is gained, but also a base of knowledge is created that can be of value for assessing the effectiveness of existing therapeutic approaches or developing new ones (Latash and Anson 1996).

Summary

Introduction

The issues addressed in this thesis concern the control of equilibrium in a bimanual, whole-body lifting task. This, for most humans daily, task comprises the forward bending of the trunk and the lowering of the whole body to grasp an object with two hands, followed by the lifting of the object to waist or chest level (Figure 1). The performance of this multi-joint movement is likely to entail a disturbance of the whole-body's equilibrium, defined as a horizontal[3] displacement of the centre of mass (CoM) of the whole body with respect to the base of support (Gahery and Massion 1981; Massion 1992). Equilibrium maintenance is, in the first place, challenged by the forward and backward bending of the upper body, inducing, respectively, a considerable forward and backward displacement of the CoM of the upper body and, without adequate postural adjustments, of the total body.

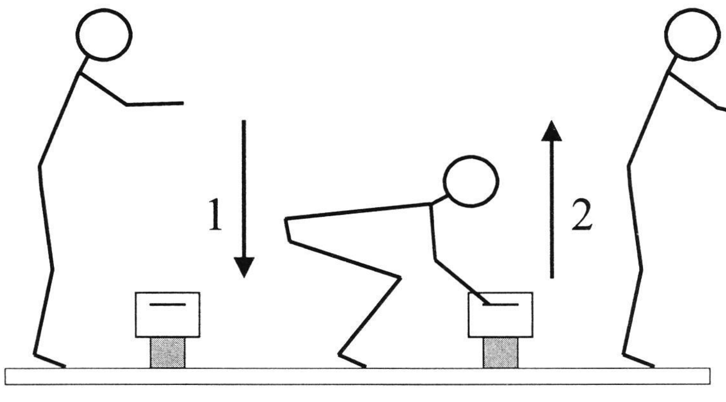


Figure 1
The investigated bimanual, whole-body lifting task. The unloaded downward movement (1) is followed by a loaded upward movement (2). The subject grasps and lifts the load in an ongoing motion.

The second challenge to equilibrium maintenance is inherent in the pick-up of an object in front of the body. The addition of an extra mass (anterior) to the body causes the CoM, of the combined body and object, to shift forward with respect to the base of support and, thus, presents the lifter with an expected perturbation to equilibrium. Furthermore, the forward

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[3] *given the sagittal plane description of the lifting task in this thesis, the horizontal direction refers to the anterior-posterior direction*

shifting CoM induces a sharp decrease in the moment exerted by the ground reaction force about the CoM (this will be elucidated below). Given the fact that a rather large moment is required to promote the whole-body extension at this stage, the forward CoM shift could impede a smooth extending motion of the whole body if not adequately anticipated.

Previous studies on the control of equilibrium have mainly investigated tasks with a *static* character, i.e. *upper-limb or upper-body movements performed in stance* (Belen'kii et al. 1967; Bouisset and Zattara 1987; Crenna et al. 1987; Massion et al. 1993). The results generally suggested that maintaining the horizontal position of the body CoM with respect to the base of support against expected or unexpected disturbances is the central rule that governs task organization (Massion 1992; 1994). Furthermore, it is commonly accepted that the ground reaction force vector should be continuously aimed at the body CoM (Murray et al. 1967; Nashner and McCollum 1985). Moreover, static tasks were found to be characterized by a clear distinction in the segments that perform the voluntary or intended movement (upper limbs or upper body) and the segments that support the body and perform the postural adjustments (lower limbs). Investigating the control of equilibrium in a static task is limited, however, in the sense that in daily life expected or unexpected equilibrium disturbances are often encountered in a *dynamic* situation, for example, while walking, rising from a seat, kicking a ball or lifting a large and heavy box. In those *whole-body movements*, the legs are involved in both the voluntary movement and the postural adjustments, which means that these postural adjustments have to be integrated in the ongoing motion of the whole body. Furthermore, for a dynamic task such as lifting and lowering a load, Toussaint et al. (1995) found that during the course of action the ground reaction force vector pointed substantially in front of or behind the CoM. Hence, the ground reaction force exerted a moment about the CoM, defined as *the external moment*. Toussaint et al. (1995) showed that the presence of an external moment did not disturb equilibrium. Rather, it was required to accomplish the segmental rotations necessary to reach the task goal, the external moment being equal to the rate of change of the whole-body angular momentum. Thus, in dynamic tasks, control of the ground reaction force vector is such that it points away from the CoM whenever necessary (to establish a change in angular momentum), whereas in static tasks, control of the ground reaction force vector is such that pointing away from the CoM is minimized.

Given the mentioned distinctions between static and dynamic tasks, the conclusions drawn with respect to the control of equilibrium in static tasks might not be simply applicable to the control of equilibrium in tasks of a dynamic nature. It may be questioned whether the regulation of the horizontal position of the CoM can be regarded as the leading principle in the organization of postural adjustments in these dynamic tasks. Would control of the CoM position suffice to maintain dynamic equilibrium and enable an undisturbed motion? Are additional principles or constraints perhaps operative to control equilibrium in a dynamic task? The present thesis aspired to answer these questions, by investigating the control of equilibrium in the above mentioned bimanual, whole-body lifting task. A two-dimensional biomechanical analysis was applied to study the integration in the primary multi-joint movement of the postural actions that counteract the equilibrium disturbances associated with the performance of the lifting task. In this analysis the lifter's motions, the underlying muscle activity patterns, the net leg joint torques, the external forces and the moments of the external forces, all in the sagittal plane of motion, were collectively examined to infer basic mechanical and/or neural principles about and constraints in the control of equilibrium. The global coordination pattern that characterizes the lifting task was described by the magnitude, direction and point of application of the ground reaction force, the external moment of the ground reaction force about the CoM, the linear CoM momenta and the whole-body angular momentum. To elucidate the differences in global patterns between various lifting tasks, these global patterns were related to local muscle activity patterns and leg joint torques. In applying this biomechanical analysis it was assumed that the lifter controls the external ground reaction force during task execution, following the work of Ingen Schenau and colleagues (Ingen Schenau et al. 1992; 1995a). These authors demonstrated, in a variety of tasks, that the actor controls the direction of an external force by regulating torques at leg joints (Doorenbosch and Ingen Schenau 1995; Ingen Schenau et al. 1992; Jacobs and Ingen Schenau 1992a). Moreover, the observed leg muscle activity patterns could be completely understood in the light of the requirement to control the direction of the external force. A similar approach was applied by Toussaint and co-workers in studying coordinative aspects of bimanual, whole-body lifting tasks (Toussaint et al. 1992; 1995).

Associated postural adjustments in a bimanual, whole-body lifting task

The possible equilibrium disturbance associated with the forward and backward bending of the upper body during, respectively, the downward approach and upward lifting phase was found to be counteracted by a specific synergy between the motions of the upper body and lower limbs (chapter 3). The forward displacement of the upper body during the downward phase was accompanied with a backward displacement of the pelvis, defined as an *associated postural adjustment*. As a result, the forward displacement of the whole-body CoM remained small, despite the large forward shift of the upper-body CoM. The multi-joint movement appeared to be coordinated such that the horizontal position of the CoM was regulated with respect to the base of support, though not maintained at a certain position. Investigation of the angular changes at the ankle, knee and hip joints suggested that the postural adjustments related to object pick-up were regulated primarily in the ankle joint and not in the knee and hip joints. Since it was assumed that the ankle, knee and hip joints were all involved in both the voluntary lifting movement and the postural adjustments, it was further investigated whether muscles crossing the ankle joint played a more pronounced role with regard to the associated postural adjustments than muscles at the other leg joints (chapter 4). The lifting task was executed on a base of support that was reduced to 0.092 m in the anterior-posterior direction (a beam). It was expected that this would augment the demands on equilibrium control and, thus, create a situation in which the hypothesized role of the ankle muscles in the control of associated postural adjustments would emerge more clearly. The kinematic synergy that was revealed during lifting on the normal base of support was in essence maintained during task performance on the beam. However, adjustments in range of motion of the ankle joint resulted in an adequate and effective adaptation of the horizontal trajectory of the CoM with respect to the reduced base of support. Less flexion in the ankle joint yielded a more backward positioned pelvis and, thus, the CoM projection remained close to the middle of the beam throughout the movement. This was accomplished by a substantial change in the electromyographic (EMG) pattern of muscles crossing the ankle joint, while muscles crossing the knee and hip joints did not show such major changes in EMG pattern. It was concluded that, although the angular changes at the ankle, knee and hip joint served both a focal and a postural role in this task, the associated postural adjustments were primarily regulated at the ankle joint.

Anticipatory postural adjustments in a bimanual, whole-body lifting task

The possible equilibrium disturbance associated with the pick-up of an object in front of the body was found to be counteracted by changes in kinematics, kinetics and leg muscle activity patterns prior to the onset of the disturbance, that is before exerting an upward force on the object (chapters 5 to 7). The anticipatory changes were identified by comparing the kinematics, kinetics and EMG patterns of a loaded downward movement, after which a load was picked up and lifted, with those of an unloaded downward movement, after which no load was picked up and lifted. The differences between both downward motions were defined as *anticipatory postural adjustments*. The anticipatory adjustments were characterized by an increase in the backward directed horizontal CoM momentum and a decrease in the forward directed (i.e. flexing) whole-body angular momentum. Thus, the adjustments appeared to be aimed at counteracting the equilibrium-disturbing effect of, respectively, the forward CoM shift (due to the addition of mass in front of the body) and the deceleration of the backward directed (i.e. extending) whole-body rotation (due to the sharp decrease in the moment exerted by the ground reaction force about the -forward shifted- CoM). The anticipatory changes in horizontal and angular momenta were accomplished by a combination of anticipatory changes in the ground reaction force: an increase in the backward directed horizontal component of the ground reaction force and a forward shift of the point of application of the ground reaction force (the centre of foot pressure). These anticipatory changes in direction and point of application of the ground reaction force were shown to be related to pronounced anticipatory adjustments in muscles crossing the ankle joint. Thus, again (see chapter 4), the ankle joint and muscles crossing this joint were found to play a special part with regard to control of postural adjustments in the bimanual, whole-body lifting task. This finding supports the assumption that the subject controls the external ground reaction force during task execution, as demonstrated in studies of Ingen Schenau and co-workers (Ingen Schenau et al. 1992; Jacobs and Ingen Schenau 1992a) and proposed in chapter 2. Whereas in the studies of Ingen Schenau and co-workers the direction of the ground reaction force was the main parameter controlled by the actor, in the present thesis the point of application of the ground reaction force was the main variable under active control of the lifter.

The anticipatory postural adjustments in kinematics and kinetics were shown to depend on the technique applied to lift the object, although it could have been expected that picking up and lifting the same load at equal speed would yield a similar perturbation in each condition (chapters 5 and 6). Apparently, the anticipatory adjustments were modulated according to the dynamic requirements of each lifting technique, taking into account that picking up and lifting the same load at equal speed has a different mechanical effect on the primary motion when lifting with different techniques. The anticipatory postural adjustments in kinematics and kinetics were furthermore shown to depend on the subjects' expectations of the load to be lifted (chapters 6 and 7). The kinematics and kinetics of the downward movement were not different prior to object pick-up in case subjects lifted a 16 kg-box which they expected to be 16 kg and in case subjects lifted a 6 kg-box which they also expected to be 16 kg. In the latter condition, the kinematics and kinetics of the downward movement were significantly different, though, from those associated with lifting a 6 kg-box, which was correctly expected to be 6 kg. The effect of load expectation on anticipatory postural adjustments was not only manifested in the parameters that quantify the primary motion (like linear and angular momenta), but also in the parameter that quantifies the mechanical load on the lumbar spine, i.e. the net torque at the lumbo-sacral joint (chapter 6). The peak torque at the lumbo-sacral joint was found to be largely determined by the *expected* load rather than the *actual* load. This means that lifting a light load, which is expected to be heavy, increases the peak mechanical load on the lumbar spine to the level associated with actually lifting the heavy load. In conclusion, the findings of chapters 5 to 7 suggest that the anticipatory postural adjustments, counteracting the equilibrium perturbation associated with picking up and lifting a load, are specified in advance on the basis of the expected load and taking into account the mechanical interaction between load pick-up and primary movement.

Control of the amount of whole-body rotation prevails in the postural adjustments in the bimanual, whole-body lifting task

In chapters 3 and 4, the regulation of the horizontal position of the CoM was regarded as the leading principle in the organization of the associated postural adjustments in the bimanual, whole-body lifting task, as was earlier proposed for upper-body movements performed in stance. Using the traditional approach to the investigation of equilibrium control (analysis of joint and body CoM translations with respect to the base of support and of leg muscle activity patterns), these chapters indeed provided results which supported the above principle. However, applying the biomechanical approach to investigate the control of equilibrium from a dynamic perspective and acknowledging the importance of segmental and whole-body rotations, chapters 5 and 6 and several other studies of our group (Toussaint et al. in press 1997a; in press 1997b) suggested that additional rules or constraints may govern the postural adjustments in the whole-body lifting task. To further identify these rules and constraints, chapter 7 (re-)investigated the anticipatory postural adjustments counteracting an expected equilibrium perturbation and the postural responses compensating an unexpected perturbation. The global mechanical analysis of the postural adjustments demonstrated that regulating the horizontal position of the CoM with respect to the base of support was not the leading cue in the organization of these adjustments. Instead, regulation of the moment exerted by the ground reaction force about the CoM appeared to take precedence, while positioning of the CoM seemed to be subordinated to this. As the external moment regulates the whole-body angular momentum, it is proposed that control of the amount of rotation of body segments, in particular of the trunk/head segment, is the leading principle in the postural adjustments during lifting.

Samenvatting

Inleiding

Een vloeiende en succesvolle uitvoering van vele dagelijkse taken vergt naast een adequate sturing van de doelgerichte handeling, ook een adequate regulatie van het evenwicht van het gehele lichaam. De regulatie van het evenwicht treedt duidelijk op de voorgrond bij het uitvoeren van taken op een klein of glad steunvlak, zoals het reiken naar een voorwerp boven het hoofd, staand op een krukje. De regulatie van het evenwicht openbaart zich ook in de reacties die volgen op een onverwachte verstoring van het evenwicht, bijvoorbeeld wanneer het voertuig waarin men staat plotseling remt. De evenwicht-regulerende reacties, zoals het maken van een stap voorwaarts of het zoeken van extra steun met de handen, zijn dan gericht op het voorkomen van een val. Het actief reguleren van het evenwicht is veel minder evident tijdens het uitvoeren van allerlei dagelijkse activiteiten, waarbij geen externe factoren te onderkennen zijn die het bewaren van het evenwicht bemoeilijken of zelfs verstoren. Echter, het vloeiend en succesvol uitvoeren van vele taken in stand of tijdens het lopen, zoals het openen van een deur, het aanpakken van een vol dienblad, het bedienen van een machine, het voorttrekken van speelgoed op wieltjes of het verzorgen van een patiënt in bed, vereist wel degelijk een actieve regulatie van het evenwicht (Massion 1992). Immers, het uitoefenen van een kracht op de omgeving, het (laten) toevoegen van extra massa aan het lichaam of het verplaatsen van lichaamsdelen met een relatief grote massa (romp of armen) ten opzichte van het steunvlak, impliceert een verstoring van het evenwicht. In navolging van Gahery en Massion (1981) en Massion (1992), wordt een verstoring van het evenwicht in dit proefschrift gedefinieerd als een horizontale[4] verplaatsing van het lichaamszwaartepunt (LZP) ten opzichte van het steunvlak .

De regulatie van het evenwicht is in het verleden met name onderzocht in 'statische' taken, taken waarin bewegingen van romp en/of armen uitgevoerd worden in stilstand, zonder bewegingen van de benen (Belen'kii e.a. 1967; Bouisset en Zattara 1987; Crenna e.a. 1987; Massion e.a. 1993). Men concludeerde dat het handhaven van de horizontale positie van het LZP ten opzichte van het steunvlak de centrale eis stelt waarop de gehele beweging wordt afgestemd (Massion 1992; 1994). Tevens bestaat er consensus over het gegeven dat de grond-reactiekracht vector tijdens stil staan voortdurend in de richting van het LZP dient te

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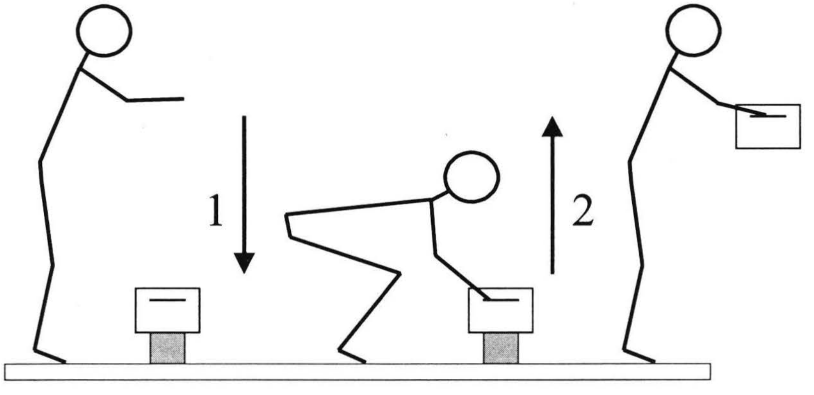
[4] *aangezien de uitvoering van de bi-manuele tiltaak beschreven en geanalyseerd wordt in het sagittale vlak, kan men de 'horizontale richting' gelijkstellen aan de 'voor-achterwaartse richting'*

wijzen (Murray e.a. 1967; Nashner en McCollum 1985). Bovendien blijkt bij statische taken een vrij strikt onderscheid gemaakt te kunnen worden tussen enerzijds de lichaamsdelen die betrokken zijn bij de uitvoering van de doelgerichte beweging (romp en/of armen) en anderzijds die delen welke het lichaam ondersteunen en betrokken zijn bij de regulatie van het evenwicht (benen).

De resultaten van het onderzoek naar de regulatie van het evenwicht in een *statische taak* zijn echter beperkt toepasbaar, aangezien verwachte of onverwachte verstoringen van het evenwicht in het dagelijks leven vaak optreden in een *dynamische situatie*, bijvoorbeeld tijdens het lopen, het opstaan van een stoel, het schoppen tegen een bal of het optillen van een zwaar object van de grond. In dergelijke taken zijn de benen altijd bij zowel de uitvoering van de doelgerichte handeling (het lopen bijvoorbeeld), als de regulatie van het evenwicht betrokken. Dit impliceert dat de evenwicht-regulerende activiteiten van de benen geïntegreerd dienen te worden in en afgestemd dienen te worden op de doelgerichte bewegingen van het gehele lichaam. Daarnaast is gevonden dat de grond-reactiekracht vector niet voortdurend in de richting van het LZP wijst tijdens een dynamische taak, zoals het optillen en neerzetten van een last (Toussaint e.a. 1995). In bepaalde fases van de taakuitvoering bleek de grond-reactiekracht vector ruim voor of achter het LZP gericht te zijn. Er was dus een moment van de grond-reactiekracht aanwezig ten opzichte van het LZP, gedefiniëerd als het *extern moment* (Toussaint e.a. 1995). De aanwezigheid van een extern moment bleek niet gepaard te gaan met een verstoring van het evenwicht. Sterker nog, de aanwezigheid van een extern moment bleek essentiëel om rotaties van lichaamsdelen (bijvoorbeeld de romp) te bewerkstellingen welke nodig zijn om de taak te kunnen uitvoeren. Immers, het impulsiemoment (de hoeveelheid rotatie) van het gehele lichaam ten opzichte van het LZP kan alleen veranderen als er een moment van een externe kracht ten opzichte van het LZP aanwezig is (Toussaint e.a. 1995, naar Greenwood 1965). De aanwezigheid van een extern moment impliceert een belangrijk onderscheid in (de sturing van) statische versus dynamische taken. In dynamische taken zal de grond-reactiekracht vector actief van het LZP af gericht worden, wanneer dat nodig is om het impulsiemoment van het gehele lichaam te veranderen, terwijl statische taken zodanig gecoördineerd worden dat afwijkingen van de grond-reactiekracht vector ten opzichte van het LZP geminimaliseerd worden.

Gezien de boven geschetste verschillen tussen statische en dynamische taken lijkt het niet vanzelfsprekend dat de conclusies met betrekking tot evenwichtsregulatie in statische taken zonder meer van toepassing zijn op dynamische taken. Het is de vraag of de handhaving van de horizontale positie van het LZP ten opzichte van het steunvlak de centrale eis stelt waarop de gehele (dynamische) beweging afgestemd wordt. Is de regulatie van de LZP-positie wel voldoende om het evenwicht te bewaren en tegelijkertijd een ongestoord bewegingsverloop te waarborgen? Zijn er misschien andere principes van toepassing op de evenwichtsregulatie in een dynamische taak? Om deze vragen te beantwoorden is in dit proefschrift de regulatie van het evenwicht in een dynamische, *bi-manuele tiltaak* bestudeerd met behulp van een biomechanische analyse. De bewegingen van de tiller, de netto gewrichtsmomenten, de onderliggende spieractivatiepatronen, de externe krachten en de momenten van externe krachten zijn gezamenlijk onderzocht, teneinde basale neurale en/of mechanische principes en/of randvoorwaarden met betrekking tot de regulatie van het evenwicht in een dynamische taak te identificeren. De globale taak karakteristieken zijn beschreven in termen van grootte, richting en aangrijpingspunt van de grond-reactiekracht, extern moment van deze kracht, LZP-impuls en impulsiemoment van het gehele lichaam. Om de herkomst van verschillen in coördinatie tussen diverse tiltaken te verklaren zijn de globale kenmerken gerelateerd aan lokale spieractivatiepatronen en netto gewrichtsmomenten. In deze benadering wordt verondersteld dat de tiller actief de grond-reactiekracht vector reguleert tijdens het tillen om de gewenste globale patronen te realiseren. Deze veronderstelling neemt een centrale positie in het proefschrift in en is gebaseerd op het werk van van Ingen Schenau en collegae (Ingen Schenau e.a. 1992; 1995a; Jacobs en Ingen Schenau 1992a).

Het oppakken en verplaatsen van relatief grote of zware voorwerpen is voor veel mensen een dagelijkse handeling. Deze bi-manuele tiltaak omvat achtereenvolgens (1) het voorover buigen van het bovenlichaam en het naar beneden brengen van het bekken, (2) het vastpakken van het te tillen voorwerp met twee handen en (3) het uitstrekken van het bovenlichaam en het opwaarts verplaatsen van het bekken om het voorwerp naar middel- of borsthoogte te transporteren (zie Figuur 1). Tijdens het uitvoeren van deze sequentie van handelingen wordt het evenwicht op twee manieren verstoord. Een eerste verstoring treedt op tijdens het voorover buigen en uitstrekken van het bovenlichaam. Dit gaat gepaard met een aanzienlijke, respectievelijk, voor- en achterwaartse verplaatsing van het zwaartepunt van het bovenlichaam èn een voor- en achterwaartse verplaatsing van het LZP ten opzichte van het steunvlak, tenzij er evenwicht-regulerende maatregelen getroffen worden. Een tweede verstoring van het evenwicht is gelegen in het oppakken van het voorwerp en de daaraan verbonden toevoeging van een extra massa aan het lichaam. Aangezien de extra massa ventraal van het LZP wordt opgepakt, zal de positie van het zwaartepunt van het totale systeem (lichaam en last) nà oppakken ventraal gelegen zijn ten opzichte van de positie van het LZP vóór oppakken. Met andere woorden, het zwaartepunt ondergaat in korte tijd (±100 ms, zie hoofdstuk 7) een aanzienlijke voorwaartse (ventrale) verschuiving (±38 mm, zie hoofdstuk 3) en vormt daarmee voor de tiller een verwachte verstoring van het evenwicht. De voorwaartse verschuiving van het systeem-zwaartepunt bewerkstelligt tevens een sterke reductie in het extern moment van de grond-reactiekracht ten opzichte van het systeem-zwaartepunt. Aangezien in dit stadium van de beweging (net na het oppakken van de last) een betrekkelijk groot extern moment vereist is om de snelheid waarmee het lichaam zich uitstrekt (het impulsiemoment) te vergroten, kan de voorwaartse verschuiving van het zwaartepunt dus een vloeiende uitstrekking van het lichaam belemmeren, tenzij er evenwicht-regulerende maatregelen getroffen worden.



Figuur 1
De bestudeerde bi-manuele tiltaak. De onbelaste neerwaartse beweging (1) wordt gevolgd door een belaste opwaartse beweging (2). De proefpersoon grijpt het voorwerp en tilt het op in een continue beweging.

Bewegingsgerelateerde houdingsaanpassingen in een bi-manuele tiltaak

De verstoring van het evenwicht die samenhangt met het voorover buigen van het bovenlichaam in de neerwaartse fase en het uitstrekken van het bovenlichaam in de daarop volgende opwaartse fase, bleek geminimaliseerd te worden door een specifieke synergie van bewegingen van boven- en onderlichaam (hoofdstuk 3). In de neerwaartse fase werd de voorwaartse verplaatsing van het bovenlichaam vergezeld door een achterwaartse verplaatsing van het bekken, een beweging die geclassificeerd werd als een *bewegingsgerelateerde houdingsaanpassing* ('associated postural adjustment'). Aldus bleef de voorwaartse verplaatsing van het LZP ten opzichte van het steunvlak beperkt, ondanks de grote voorwaartse verplaatsing van het zwaartepunt van het bovenlichaam ten opzichte van het bekken. Tijdens de taakuitvoering bleek wel een actieve regulatie van de horizontale positie van het LZP plaats te vinden, doch er was geen sprake van het handhaven van de uitgangspositie van het LZP ten opzichte van het steunvlak. Het oppakken van de last, het tweede type evenwichtverstoring in deze taak, bleek gepaard te gaan met een aanpassing van de rotatie van het onderbeen ten opzichte van de voet. Een dergelijke aanpassing van de hoekverandering in een gewricht werd niet gevonden in de knie- en heupgewrichten. Dit zou kunnen betekenen dat de evenwicht-regulerende aanpassingen, gerelateerd aan het oppakken van een last, met name gerealiseerd werden door spieren rond het enkelgewricht.

In hoofdstuk 4 werd een mogelijke dominante rol van de spieren rond het enkelgewricht in de evenwichtsregulatie tijdens bi-manuele tiltaken nader bestudeerd. De proefpersonen voerden dezelfde tiltaak uit terwijl ze op een balkje stonden. Aldus werd het steunvlak in voor-achterwaartse richting gereduceerd tot 92 mm. Aangezien deze conditie hoge eisen stelt aan de evenwichtsregulatie, werd verwacht dat de veronderstelde dominante rol van de enkelspieren in de organisatie van bewegingsgerelateerde houdingsaanpassingen prominenter naar voren zou treden. De specifieke synergie van bewegingen van boven- en onderlichaam, geïdentificeerd tijdens het tillen op een niet-gereduceerd steunvlak, werd ook gevonden tijdens het tillen op het balkje. Daarbij bleek de horizontale verplaatsing van het LZP ten opzichte van het steunvlak op effectieve wijze afgestemd te worden op het gereduceerde steunvlak, namelijk door een aanpassing van de rotatie van het onderbeen ten opzichte van de voet. In vergelijking met de taakuitvoering op een niet-gereduceerd steunvlak werd minder (dorsaal) flexie waargenomen in het enkelgewricht. Dit resulteerde in een achterwaartse verschuiving van het (gehele) horizontale bewegingstraject van het bekken, waardoor de projectie op het steunvlak van het horizontale bewegingstraject van het LZP rond de middellijn van het balkje fluctueerde. Deze aanpassing in LZP-traject werd gerealiseerd door een aanzienlijke verandering in het activatiepatroon van spieren rond het enkelgewricht, terwijl de spieren rond het knie- en heupgewricht relatief weinig veranderingen in activatiepatroon vertoonden. In de onderzochte bi-manuele tiltaak werd de bewegingsgerelateerde houdingsaanpassing (de compensatoire verplaatsing van het bekken) dus inderdaad vooral gerealiseerd door de spieren rond het enkelgewricht.

Anticipatoire houdingsaanpassingen in een bi-manuele tiltaak

De verstoring van het evenwicht die samenhangt met het oppakken van een last bleek geminimaliseerd te worden door veranderingen in de kinematica, in de kinetica en in de activatiepatronen van beenspieren (hoofdstuk 5, 6 en 7). Deze veranderingen werden waargenomen vóór de last werd vastgepakt en werden geclassificeerd als *anticipatoire houdingsaanpassingen* ('anticipatory postural adjustments'). Ze werden geïdentificeerd door de kinematica, kinetica en spieractivatiepatronen behorende bij een neerwaartse beweging die *niet* gevolgd werd door het optillen van een last te vergelijken met de kinematica, kinetica en spieractivatiepatronen behorende bij een neerwaartse beweging die *wel* gevolgd werd door het optillen van een last. Verschillen tussen beide neerwaartse bewegingen waren derhalve te relateren aan het (toekomstige) optillen van de last. De anticipatoire houdingsaanpassingen uitten zich in een toename van de achterwaarts gerichte horizontale impuls van het LZP en in een afname van het voorwaarts gerichte (flecterende) impulsiemoment van het gehele lichaam. Aldus reduceerden deze aanpassingen de evenwicht-verstorende gevolgen van, respectievelijk, de voorwaartse LZP verschuiving (inherent aan de toevoeging van massa ventraal van het LZP) en de remming van de rotatiesnelheid waarmee het lichaam zich uitstrekt (inherent aan de sterke reductie in het extern moment van de grond-reactiekracht ten opzichte van het systeem-zwaartepunt). De anticipatoire veranderingen in horizontale impuls en impulsiemoment werden gerealiseerd door een combinatie van anticipatoire veranderingen in de grond-reactiekracht: een toename van de achterwaarts gerichte horizontale component van de grond-reactiekracht en een voorwaartse verschuiving van het aangrijpingspunt van de kracht. Deze anticipatoire houdingsaanpassingen in richting en aangrijpingspunt van de grond-reactiekracht bleken gerelateerd te zijn aan anticipatoire aanpassingen in de activiteit van spieren rond het enkelgewricht. Dus evenals in hoofdstuk 4 bleken de spieren rond het enkelgewricht een hoofdrol te spelen in de evenwicht-regulerende houdingsaanpassingen tijdens de uitvoering van de bi-manuele tiltaak. Dit resultaat strookt met de theorie dat een persoon actief de grond-reactiekracht reguleert tijdens de uitvoering van taken waarin de uitoefening van een kracht op de omgeving centraal staat (Ingen Schenau e.a. 1992; Jacobs en Ingen Schenau 1992a). Echter, in de taken die door van Ingen Schenau en collegae zijn bestudeerd werd met name de richting van de grond-reactiekracht vector actief gereguleerd, terwijl in dit proefschrift met name het aangrijpingspunt actief gereguleerd bleek te worden door de tiller.

De grootte en aard van de anticipatoire houdingsaanpassingen in kinematica en kinetica bleken afhankelijk te zijn van de techniek waarmee de tiltaak werd uitgevoerd (hoofdstuk 5 en 6). De anticipatoire aanpassingen werden dus niet alleen beïnvloed door de te tillen massa en de bewegingssnelheid, aangezien deze gelijk waren bij het tillen met verschillende technieken. Blijkbaar werden de anticipatoire aanpassingen afgestemd op de toekomstige mechanische interactie tussen de last enerzijds en de beweging van het lichaam anderzijds. De grootte en aard van de anticipatoire houdingsaanpassingen in kinematica en kinetica bleken tevens afhankelijk te zijn van de verwachting die proefpersonen hadden omtrent de massa van de te tillen last (hoofdstuk 6 en 7). De bewegingskarakteristieken van de neerwaartse fase waren vrijwel identiek in de conditie waarin proefpersonen een doos van 16 kg optilden, terwijl ze deze massa verwachtten, en de conditie waarin proefpersonen een doos van 6 kg optilden, maar een doos van 16 kg verwachtten. De bewegingskarakteristieken van de neerwaartse fase in laatstgenoemde conditie waren echter wel significant verschillend van de karakteristieken van de neerwaartse fase in de conditie waarin proefpersonen een doos van 6 kg verwachtten te gaan tillen en deze ook daadwerkelijk gepresenteerd kregen. Het effect van subjectieve verwachting omtrent de te tillen last kwam niet alleen tot uiting in bewegingsparameters, zoals impuls en impulsiemoment, maar ook in de parameter welke de mechanische belasting op de lumbale wervelkolom weergeeft, namelijk het netto moment ten opzichte van de lumbo-sacrale gewricht (hoofdstuk 6). Het piek-moment ten opzichte van het lumbo-sacrale gewricht bleek grotendeels bepaald te worden door de *verwachte* last en niet door de *werkelijke* last. Dit betekent dat het tillen van een voorwerp met een geringe massa kan leiden tot een piek-belasting op de lumbale wervelkolom welke normaliter gevonden wordt tijdens het tillen van een voorwerp met een grote massa, indien de persoon niet de geringe, maar de grote massa verwacht te gaan tillen. In beroepen waar veelvuldig objecten getild en verplaatst worden is het dus belangrijk dat werknemers adequate informatie hebben omtrent de (wisselende) object massa.

Regulatie van de hoeveelheid van rotatie van het lichaam prevaleert in de evenwicht-regulerende houdingsaanpassingen in een bi-manuele tiltaak

In hoofdstuk 3 en 4 werd de regulatie van de horizontale positie van het LZP ten opzichte van het steunvlak opgevat als de centrale regel in de planning en uitvoering van evenwicht-regulerende houdingsaanpassingen. Immers, onderzoek had in het verleden veelvuldig aangetoond dat handhaving van een gegeven LZP-positie of herstel van een verstoorde LZP-positie dè kenmerken waren van evenwichtsregulatie tijdens bewegingen van romp en/of armen uitgevoerd in stand (Massion 1992; 1994). Gebruik makend van de traditionele onderzoeksmethoden (analyse van de translatie van gewrichten en van het LZP ten opzichte van het steunvlak en analyse van beenspieractivatiepatronen) werd in hoofdstuk 3 en 4 inderdaad ondersteuning gevonden voor de bovengenoemde centrale regel. Echter, toepassing van de 'nieuwe' biomechanische methode (analyse vanuit dynamisch perspectief, rekening houdend met de aanwezigheid van een aanzienlijke rotatie-component in de gehele beweging) in hoofdstuk 5 en 6 en in andere onderzoeken van onze groep (Toussaint e.a. in druk 1997a; in druk 1997b) leverde resultaten op die bovengenoemde centrale regel ter discussie stelden. Additionele principes leken de uitvoering van evenwicht-regulerende houdingsaanpassingen te bepalen, zoals de mechanische interactie tussen lastinertie en beweging (hoofdstuk 5 van dit proefschrift; Toussaint e.a. in druk 1997a) en de noodzaak om de mechanische belasting op de lumbale wervelkolom te beperken (Toussaint e.a. in druk 1997b). Om deze principes verder in kaart te brengen, werden in hoofdstuk 7 de houdingsaanpassingen die anticipatoir een verwachte verstoring van het evenwicht minimaliseren en de houdingsaanpassingen die reactief een onverwachte verstoring van het evenwicht compenseren (nogmaals) bestudeerd. De biomechanische analyse van de anticipatoire en compensatoire houdingsaanpassingen toonde aan dat regulatie van de horizontale LZP-positie niet centraal stond in de organisatie van deze aanpassingen. Er bleek voorrang gegeven te worden aan het vergroten (in de anticipatoire houdingsaanpassingen) of beperken (in de compensatoire houdingsaanpassingen) van het extern moment van de grond-reactiekracht ten opzichte van het LZP. De positionering van het LZP leek zelfs in dienst te staan van het genereren van het gewenste extern moment. Aangezien het impulsiemoment van het gehele lichaam gereguleerd wordt door het extern moment, kan gesteld worden dat regulatie van de hoeveelheid van rotatie van delen van het lichaam, met name van romp en hoofd, het centrale principe is in de evenwicht-regulerende houdingsaanpassingen in een bi-manuele tiltaak.

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Nawoord

Het proefschrift dat voor U ligt vormt voor mij de afsluiting van een belangrijke en enerverende periode in mijn leven. Ik heb als Assistent in Opleiding de kans gekregen om mezelf gedurende ruim vier jaar te bekwamen in het verrichten van fundamenteel wetenschappelijk onderzoek op een prachtig gebied, namelijk het menselijk bewegen. Ik heb genoten van deze periode, enerzijds door de inhoud van het werk en anderzijds door de prettige samenwerking met naaste collega's. Ik heb veel geleerd in deze vier jaren, niet alleen door het uitvoeren van mijn eigen onderzoeken en het volgen van relevant onderwijs, maar zeker ook door het meewerken aan onderzoeken van collega's, het begeleiden van stage-studenten, het meedenken en -werken aan het facultaire onderwijs, het optreden als voorzitter en woordvoerder van de FBW- en IFKB-promovendi, de paraboolvlucht-experimenten en niet te vergeten mijn verblijf in Stockholm. Graag wil ik iedereen bedanken die bijgedragen heeft aan de totstandkoming van dit proefschrift en/of aan de waardevolle invulling van deze periode. Bij een aantal mensen wil ik in het bijzonder even stil blijven staan.

Allereerst wil ik mijn co-promotor, Huub Toussaint, hartelijk danken voor de vrijwel 'alle-daagse' begeleiding. Beste Huub, je hebt een bijzonder grote rol gespeeld in dit proefschrift en in mijn leerproces tot volwassen onderzoeker. Je bent mijn aanspreekpunt geweest op vrijwel elk gebied. Altijd nam je tijd voor me en als er geen tijd was, was er in ieder geval aandacht. Ik heb veel kunnen leren van jouw manier van onderzoeken, onderwijzen en besturen en ik dank je dat je me bij zoveel meer dan alleen het onderzoek hebt betrokken. Je hebt me de gelegenheid geboden om zelf te ervaren en te leren, je hebt me op de juiste momenten de broodnodige duw in de rug gegeven en je hebt een werksfeer gecreëerd waarin ik me volledig thuis voelde. In de afgelopen jaren ben je bovendien een waardevolle vriend geworden, bedankt!

Graag wil ik mijn promotor, Gerrit Jan van Ingen Schenau, noemen. Beste Gerrit Jan, ondanks het feit dat je pas bij mijn promotie betrokken werd op het moment dat de grote lijnen al uitgezet en de experimenten reeds voltooid waren, heb je me alle vertrouwen gegeven. Dank hiervoor. Ik heb veel respect voor jouw deskundigheid, creativiteit, enthousiasme en passie voor onderzoek. Hopelijk biedt de toekomst me de gelegenheid om hiervan nog wat langer te kunnen genieten. Dank, tot slot, voor de aangename gesprekken over Zweden, schaatsen en schaatsen in Zweden.

Mijn collega 'Tillers', Jaap van Dieën, Idsart Kingma en Michiel de Looze wil ik hartelijk bedanken voor alle steun, antwoorden op mijn vragen en vragen naar mijn antwoorden, gezellige lunches, borrels en etentjes, zinnige discussies en onzinnige, maar wel leuke, conversaties. Het was bijzonder (ofwel 'buitensporig' I.K.) goed toeven in deze club en ik hoop van harte dat mijn lidmaatschap voorlopig nog niet opgezegd hoeft te worden.

Daarnaast wil ik mijn vakgroep- en/of lijngenoten, collega's van de 'Human Performance Group' (met name Peter Hollander), collega-promovendi (met name Tom Welter en Lieke Peper) en een enkeling die tot geen van bovengenoemde categorieën behoort (namelijk Geert Savelsbergh en Peter Beek) bedanken voor waardevolle discussies en adviezen, het delen van een boeiende wetenschap en/of gewoon wat gezelligheid.

Ook wil ik op deze plaats de voorzitter van de 'oude' vakgroep Gezondheidkunde, Han Kemper, bedanken voor de adviezen, begeleiding en praktische hulp bij het inspanningsfysiologische til-onderzoek waar ik mijn AiO-periode mee begonnen ben.

De technische ondersteuning van Marty van Bakel, Evert van de Belt, Jos van den Berg, Idsart Kingma, Ton Meulemans, Willem Schreurs en Frank Wijkhuizen was belangrijk voor het welslagen van de onderzoeksprojecten. Bedankt! Ook de administratieve ondersteuning van mijn activiteiten wil ik niet vergeten. Dank aan Marloes den Besten, Marion Lochmans, Arthur van der Meer, Barbara Oudejans en Olga Schipper. Tevens wil ik Janneke Kuijken bedanken voor haar steun en adviezen als AiO-mentor.

Graag wil ik Marco Hoozemans en Michiel Ober bedanken voor hun bijdrage aan de onderzoeken beschreven in hoofdstuk 5 en 6, voor hun inzet, enthousiasme, de amusante 'Avonturen van Hoos en Ober in het Tilproject' en andersoortige (Rand-)stadse grollen. Ook wil ik, ondanks het feit dat ze niet 'mijn' studenten waren, David de Louw en Jacqueline Vos bedanken voor de prettige samenwerking en de waardevolle discussies.

Geen experimenteel bewegingswetenschappelijk onderzoek zonder proefpersonen! Dank aan alle 'young, healthy male subjects' (n=50, mean age 23.0 years, mean body height 1.79 m and mean body mass 72.4 kg).

Jag skulle vilja också tacka mina svenska kollegor och vänner! Bästa Helga, du har varit och fortfarande är ett inspirerandet exempel och en mycket snäll person. Det är en stor förmån att samarbeta med dig. Bästa Lena, tack ska du ha för många underbar vackra skridskoturer (jag tacker Sten också), den trevliga samverkan och din fina vänskap. Och tack så mycket, bästa Ing-Marie, för gästfriheten och för rådet och hjälpen på labet. Jag ska aldrig glömma mina tre jätte-speciella månader i Stockholm!

I also wish to express my gratitude to dr. J. Massion (Centre National de la Recherche Scientifique Marseille), dr. H. Hirschfeld (Karolinska Institutet Stockholm), dr.ir. A.L. Hof (Rijksuniversiteit Groningen), dr. A.P. Hollander (Vrije Universiteit Amsterdam), dr. A.J. van Soest (Vrije Universiteit Amsterdam) and prof.dr. Th.W. Mulder (Sint Maartenskliniek Nijmegen) for reviewing this thesis.

Ik wil op deze plaats tevens mijn ouders bedanken, me daarbij realiserend dat ze aan veel meer mijlpalen in mijn leven dan alleen dit proefschrift een waardevolle bijdrage geleverd hebben. Lieve vader en moeder, jullie hebben me geleerd zorgvuldig om te gaan met de talenten die ik gekregen heb en jullie hebben me gesteund om die talenten zo goed mogelijk te ontwikkelen. Het is tijdens mijn (studenten-)jaren op kamers voor mij heel belangrijk geweest dat er in Rucphen altijd een basis bleef, waar ik thuis kon komen en waar ik rust had. En ook nu kom ik, alhoewel minder frequent, nog heel graag 'naar huis'. Bedankt!

Tot slot wil ik natuurlijk Maurice bedanken. Lieve Maurice, je bent een kei! Je hebt me alle ruimte gegeven om mezelf te ontwikkelen als wetenschappelijk onderzoeker, sterker nog, je hebt me daarin altijd gestimuleerd. Door jouw begrip heeft mijn liefde voor de wetenschap zelden hoeven te wedijveren met mijn liefde voor jou. Ik ben bijzonder blij en dankbaar dat je trots bent op mijn werk en dit proefschrift.

About the author

Dianne (Dimphena, Adriana, Cornelia, Maria) Commissaris was born on the 10th of November 1966 in Rucphen (Noord-Brabant) and grew up, with four brothers, in a catholic family. From 1979 to 1985 she attended the 'Gertrudis Lyceum' in Roosendaal and passed her final exam (Gymnasium-β) with credit. The next four years she attended the 'Hogeschool Heerlen', department of Occupational Therapy in Hoensbroek. After graduation in June 1989, she started her study at the Faculty of Human Movement Sciences, Vrije Universiteit Amsterdam and graduated with credit in August 1992 with a major in Health Sciences and a minor in Psychology. In September 1992, she commenced to work at the Faculty of Human Movement Sciences in Amsterdam, the first few months as a research assistant and from December 1992 to March 1997 as a PhD-student at the Amsterdam *Spine* Unit ('TilProject'). She first investigated the mechanics and energetics of repetitive lifting and started in October 1993 with the research on the control of equilibrium in bimanual, whole-body lifting tasks, which resulted in the present PhD-thesis. In the (lovely skating) winter of 1996, she worked for three months at the lab of dr. Hirschfeld, at the Karolinska Institute in Stockholm. She conducted her most thrilling experiment in December 1996, under zero-gravity conditions during the 23rd ESA Parabolic Flights Campaign on board the NASA KC-135 above the Gulf of Biscay. As a PhD-student, she was chairman of the association of PhD-students of the Faculty of Human Movement Sciences for one and a half year. For many years, she spent quite some spare time working with mentally retarded and physically handicapped children, disabled adults and elderly and non-disabled teen-agers. Sport has always been her most important leisure activity, varying throughout the years from all kind of secondary school tournaments to long-distance running, sailing, indoor soccer, ice-skating, cycling and fitness.