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Lower extremity movements in frontal plane balance control during one-leg stance

De bewegingen van de onderste extremiteit tijdens evenwichtshandhaving in het frontale vlak bij het staan op één been

Proefschrift

ter verkrijging van de graad van doctor
aan de Erasmus Universiteit Rotterdam
op gezag van de Rector Magnificus
Prof.dr P.W.C. Akkermans M.A.
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opgedragen aan Willemien, Cees

en mijn (groot)ouders

Contents

Chapter I

General introduction

The beginning

A major turning point in human phylogenetic development occurred with the transition of Hominidae (anthropoid humans) from four-leg stance to two-leg stance. There are two different theories about the cause as well as the time of this transition. A widely adopted view is that the transition from four-leg stance to two-leg stance occurred about 3.5 million years ago when, as the result of the increased brain mass, the capacity for the construction and utilization of tools for hunting and food preparation was obtained. In the absence of fossils to support this theory, an alternative view was developed according to which the transition occurred much earlier, about 10 million years ago, due to the altered social and behavioral patterns[1]. Whatever the cause of its initiation, the adoption of biped stance and locomotion must have had distinct advantages over quadruped stance for the survival of these early humanoids.

In contemporary western society the power to adopt and maintain the erect position is no longer a prerequisite for survival. Nevertheless, our society is so strongly geared towards this capacity that its absence can be a major burden. In ontogenetic development the capability to adopt biped stance arises about 10-15 months after birth. Hereafter the power to adopt biped ambulation and one-leg stance develop successively. A decreased capability of biped stance and ambulation can be of a congenital nature, e.g. cerebral palsy, muscular dystrophies and developmental coordination disorder, or diseases, e.g. multiple sclerosis and cerebrovascular accidents, and trauma later in life. Because adequate postural control establishes a stabilizing framework and support for manipulative skills, its absence can also hamper upper extremity movement patterns. When decreased postural control occurs at an early age this has the additional disadvantage of impairing the development of these movement patterns. In all these cases it is important, sometimes for diagnosis, but more often for the assignment, initiation and cessation of therapy to have a method to assess the coordination of the movements of the lower extremities. In the absence of such a method it was decided to pursue the development of it. An extra prerequisite was that, considering the developmental aspects, the method should be applicable in a juvenile rehabilitation context and therefore be non-invasive. In view of their importance for the preservation of erect posture during many of the activities of daily living (ADL), it seemed logical to assess the movements of the lower extremities during balance control.

Balance research

Nowadays, it is widely accepted that the study of balance and postural control has considerable clinical relevance in view of the importance of biped stance and ambulation for

ADL. A milestone in balance and postural control research are the observations of the German neurologist Moritz Heinrich Romberg (1795-1873), of increased postural sway in patients with tabes dorsalis. Romberg examined the integrity of the neuromuscular system of these patients by testing their ability to stand without support with the feet together and with open and closed eyes. He then visually assessed the amount of body sway. When the subject staggers or falls the test is said to be positive[2]. However, the interest in neuromuscular functioning has not always been the main reason for the examination of balance and posture. In the beginning of the twentieth century it was believed that "man was a made-over animal"[3] and that the altered "gravitational stresses" due to the adoption of biped stance were associated with visceral malpositions and as such the cause of many illnesses. Later, this view was abandoned and balance research was focused again on the system governing the control of balance and posture. Despite its seniority the Romberg test remained an important method for the clinical assessment of the functioning of the latter system. It is clear, however, that the interpretation of the Romberg test strongly depends on the experience of the observer. Also, this test allows ample room for differentiated assessment of balance. As a result of these limitations the need for a more objective and refined assessment of balance arose. To obtain a more quantitative description of body sway, subjects were placed in front of grid patterns[4], head movements were registered[5, 6, 7, 8], and subjects were placed on platforms that respond to foot to ground forces[9, 10]. These platforms are the predecessors of the contemporary forceplate, which consists of a platform supported by four or, preferably, three[11, 12] force transducers. From the measured forces, several variables can be calculated that are related to the movements of the subject standing on the platform and are used to describe and quantify postural control.

It is generally accepted that the system governing balance and postural control consists of four sub-systems, i.e. the visual, vestibular, proprioceptive and musculoskeletal system[13, 14, 15]. The first three systems provide the input, while the latter provides the motor reaction. Because unperturbed two-leg stance is a relatively mild challenge for the system governing postural control, it is less suitable to discover the more subtle defects in the postural control system. To overcome this problem, several modifications of standard forceplate measurements have been developed with the object to increase the demands upon the postural control system. Among these things are: eye closure[16] and visual field manipulation[17] to manipulate the visual input, caloric stimulation[18, 19] to manipulate the vestibular input, local anesthetics[20], ischemia[21] and foam surfaces[17, 22-30] to manipulate the proprioceptive system. Modifications that more specifically affect the motor system are moving[31, 32] or tilting[33, 34] forceplates, and mechanical pushes[35, 36]. A potential problem in the assessment of the functioning of the postural control system is the amount of attention devoted to the balancing task. When the amount of attention devoted to the balancing task is decreased by the simultaneous execution of a second task this results in deteriorated postural control[37, 38]. Because the amount of attention is difficult to quantify, attentional differences are a potential source of intra- and inter-subject variability in forceplate measurements.

A vast amount of work has been done on balance and posture control during two-leg stance, especially in the sagittal plane. A very important issue is the conclusion that there are basically two mechanisms for sagittal plane postural control: namely, the ankle strategy and the hip strategy[39]. In the ankle strategy the body primarily moves as an inverted pendulum about the ankle joint. Compensatory moments of force are generated about the ankle joint by the muscles in the shank. In the hip strategy compensatory, horizontal shear forces are produced through rotation of the upper body about the hip joint. Being able to distinguish these mechanisms considerably increased the insight in balance control.

Aim of study

It is widely recognized that the lower extremities play a major role during balance control in erect stance. However, despite the vast amount of research conducted on balance control, the relation between lower extremity movements and balance control in the frontal plane remained unclear. This resulted in an incorrect interpretation of the results of forceplate assisted balance research. This thesis attempts to correct this omission by establishing a theoretical framework incorporating the movements of the lower extremities in relation to their task during the maintenance of the upright position. Subsequently, this framework is used for the development of a method to assess the coordination of the lower extremities.

It was decided to study lower extremity movements during one-leg stance. In the first place, we believed that the lower extremity movements that are of major importance in balance control during one-leg stance play the same role, although less prominent, during walking. Because of this ubiquitousness, the functional integrity of this mechanism is important for many activities during daily living. In the second place, it was our opinion that there were lacunae in the existing knowledge about frontal plane balance control on one-leg stance. In addition we had alternative ideas about the contribution of lower extremity movements to frontal plane balance control. Finally, one-leg stance has the additional advantage that it poses a greater challenge for the system governing postural control. This is necessary because it is not uncommon for diseases that strongly affect motor behavior at a later age to show only mild manifestations in fine movements early in life. At an early age motor behavior will be within established reference parameters up to a considerable performance level. It is only beyond this level that the system fails to meet the demands made on it. A additional disadvantage of measurements during two-leg stance is that this position is relatively devoid of fine, highly coordinated movements. Therefore we confined our study to the frontal plane and one-leg stance. In the subsequent chapters we describe how we approached the goal of this study.

Chapters

In chapter 2 we present and validate a model that establishes a link between a movement of the lower extremity and variations in a forceplate variable.

In chapter 3, on the basis of the model described in chapter 2, we discuss the effect of the compliance of the supporting surface on the efficiency of the movements of the lower extremity with respect to their contribution to balance control.

In chapter 4, on the basis of the model described in chapter 2, we propose and test the intra-subject effect of foot breadth on the efficiency of the movements of the lower extremity with respect to their contribution to balance control.

In chapter 5, based on the model described in chapter 2, we hypothesized a negative effect on balance control of ankle bracing via a constraining effect on the movements of the lower extremity. This hypothesis was subsequently tested.

In chapter 6 we examined the combined effect of ankle bracing and decreased compliance of the supporting surface on balance control.

In chapter 7 we re-assessed the relation between foot size and the efficiency of the movements of the lower extremity with respect to their contribution to balance control in an adult and a juvenile population. Subsequently, we examined the effect of increased compliance of the supporting surface on the efficiency of the movements of the lower extremity with respect to their contribution to balance control as a function of the pressure under the foot.

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Chapter II

A model for the relation between the displacement of the ankle and the center of pressure in the frontal plane, during one-leg stance

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List of abbreviations

CoP	center of pressure
CoG	center of gravity
EQ	the efficiency quotient of foot tilt
$b_{f\text{-}max}$	maximum foot breadth
FTS	foot tilt strategy, a mechanism for balance control
HAT	body segment comprising of head, arms and trunk
RSI	rocker shaped interface with the supporting surface, representing the supporting foot
φ	angle of frontal plane foot tilt
α	angle of the segment between the hip joint and the subtalar joint, relative to the working line of the force of gravity.
h_M	the maximum height of the lateral malleolus
l_l	length of the supporting leg
M	the horizontal, frontal plane position of the lateral malleolus
R	the radius (mm) of the circle of which the curved surface of the RSI is a part
U	the horizontal, frontal plane position of the upper body
X	the frontal plane position of the CoP relative to the forceplate coordinate system
Y	the sagittal plane position of the CoP relative to the forceplate coordinate system
F_n	the horizontal ground reaction force in the frontal (n=x) or the sagittal (n=y) plane
F_g	the force of gravity acting on the body's CoG
F_{rg}	the vertical ground reaction force acting on the supporting foot

INTRODUCTION

Balance control has been a topic of interest in clinical medicine since Romberg's observations in the previous century of increased postural sway in patients with tabes dorsalis. Since then, much research has focused on balance and posture control. A popular method to assess balance is stabilometry or posturography. In this method a forceplate, i.e. a platform supported by force transducers, measures the vertical, fore-aft and medio-lateral components of the ground reaction force vector. Initially, in stabilometry, the center of pressure (CoP), defined as the point of application of the resultant vertical reaction force vector, was taken to be the projection of the center of gravity (CoG) on the supporting surface[1, 2]. Later, however, this was shown to be correct only in the absence of horizontal accelerations and movements[3-6]. Because, during two- and especially during one-leg stance, movements and consequently forces and accelerations in the horizontal plane are omnipresent, in these situations interpretation of the CoP as the vertical projection of the CoG is erroneous. Despite the resulting uncertainty as to what is actually being measured through the CoP, during one-leg stance the amplitude of its movements is often used as a measure of body sway[7-16]. Because displacement of the CoP relative to the CoG contributes to balance control in the sagittal[17] as well as the frontal plane[18], it would facilitate the interpretation of variables based on CoP displacement if the source of this displacement in terms of movements of (a part of) the body were known.

In the process of establishing a link between CoP displacement and the movements of (a part of) the body we used the equations for the calculation of the CoP, as given by Kistler[19] as a starting point. From these equations it can be deducted that the position of the CoP depends on the distribution of the total vertical force over the individual forceplate transducers and, therefore, on the position of the foot as well as the pressure distribution under the foot relative to the forceplate coordinate system. Because in stabilometry the position of the foot on the forceplate is fixed, changes in the pressure distribution under the foot are assumed to be the single source of CoP displacement. Tilting of the foot will result in considerable changes in this pressure distribution and, if present, can therefore be expected to be a major source of CoP displacement.

Since clearly discernible tilting movements of the foot predominantly occur in the frontal plane, we will further restrict ourselves to frontal plane balance control. When the position of the foot on the forceplate is fixed, a change in frontal plane foot tilt ($\Delta\varphi$) will result in a horizontal, frontal plane displacement of the lateral malleolus (ΔM) in the same direction. Unless stated otherwise, the variations in all variables of interest are considered relative to their average values. We take ΔM as a measure for $\Delta\varphi$ because this has the advantage of allowing measurement of foot tilt even when the foot is covered, e.g. by footwear. In the future this will enable the assessment of the effect of various types of footwear on the relation between foot tilt and CoP displacement and on balance control. Developing a model for the relation between frontal plane foot tilt and CoP displacement will aid in the assessment of the effects of anthropometric (foot size) or external (footwear or orthoses) factors on balance control. Ultimately such a model could be incorporated in dynamic modelling of balance control during one-leg stance.

The aim of this study is to propose and validate a model for the relation between foot tilt and CoP displacement in the frontal plane. Because this model is assumed to be linear, we assessed the linearity of the relation between frontal plane CoP displacement (ΔX) and ΔM. In order to be able to put the relation between ΔX and ΔM into perspective, and to compare our findings with those reported by others, we also assessed the relation between ΔX on the one hand and, on the other, upper body displacement (ΔU), horizontal CoG displacement (ΔCoG) and horizontal ground reaction forces (F_x) in the frontal plane, and CoP displacement (ΔY) and horizontal ground reaction forces (F_y) in the sagittal plane. Finally we will discuss some of the implications of this model for the study of balance control.

The model

The assumption that ΔX and ΔM are linked led us to abandon existing inverse pendulum models. The conventional multi-link inverse pendulum model (Figure 1A) consists of two segments representing the Head, Arms and Trunk (HAT) and the supporting leg interconnected through a link representing the hip joint. An additional link represents the ankle joint and serves the interaction with the supporting surface. For the purpose of simplicity the CoG of the body is modelled as being located in the HAT segment. All joints are modelled as hinge joints. A disadvantage of this model is that the point of interaction with the supporting surface is fixed and cannot account for displacement of the CoP. Later this model was extended by adding a third segment, representing the foot (Figure 1B), for the interaction with the supporting surface[20] and allowing displacement of the CoP. This type of foot to ground interface is predominantly used in the analysis of sagittal plane balance control. In the sagittal plane, balance is controlled by the ankle strategy and the hip strategy. In the ankle strategy the body primarily moves as an inverted pendulum about the ankle joint. This mechanism is effective with long, rigid support surfaces and displacement distances short in relation to foot length. In the hip strategy compensatory, horizontal shear forces are produced through rotation of the upper body about the hip joint[21]. A disadvantage of the type of foot model used for the ankle strategy is that it cannot account for a continuous linking of ΔM and ΔX. Instead, it allows an infinite number of angles of foot tilt to be linked with only three discrete positions of the CoP, namely in the middle of the foot when the foot rests flat on the floor (Figure 1B) and at the left or right edge of the foot when it is tilted to the left or the right (Figure 1C). Once the foot is tilted, variations in the amount of tilt do not affect the position of the CoP.

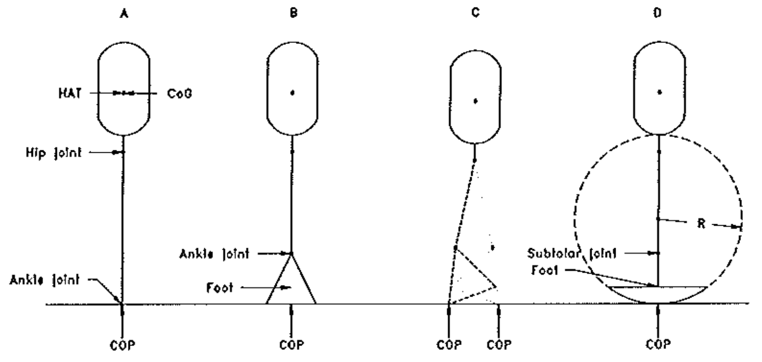

Figure 1. Inverse pendulum models: conventional (A), conventional with additional triangular foot-ground interface (B), CoP displacement and foot tilt with a triangular foot model (C), modified RSI model (D).

To accommodate this restriction, our modified model (Figure 1D) consists of a conventional, multi-link inverse pendulum model with an additional Rocker-Shaped Interface (RSI) with the supporting surface, representing the foot. The subtalar joint links the leg and the foot segment. The linked segment model, including the RSI, hereafter will be referred to as "the RSI-model". RSI tilt is consistently linked with the movements of the CoP because the point of contact between the RSI and the supporting surface is their only point of contact and therefore equals, at any time, the position of the CoP. Every angle of foot tilt is linked exclusively to one position of the CoP and foot tilt without CoP displacement is not possible. This foot-ground interface also accounts for the continuous, frontal plane ankle movements experienced during one-leg stance. The tilting movements of the RSI, relative to the shank, are caused by the contraction of muscles across the subtalar joint. It follows from the configuration of the RSI that the relation between ΔX (mm) and Δφ (radian) can be described as

$$\Delta X = \Delta\varphi \cdot R \qquad (1)$$

with R in mm as the radius of the circle of which the circumference of the RSI is a part (Figure 1D). It can be concluded from equation (1) that R determines ΔX per radian of foot tilt. The value of ΔX per Δφ in mm rad^{-1} can be regarded as the efficiency of foot tilt with respect to the generation of CoP displacement. This Efficiency Quotient of foot tilt (EQ) in mm rad^{-1} is numerically equal to R in mm. A decrease (increase) in EQ means that a certain amount of foot tilt will result in a smaller (larger) displacement of the CoP and that more (less) foot tilt is necessary to generate the same amount of CoP displacement. From equation (1) it follows that

$$\Delta\dot{X} = \Delta\dot{\varphi} \cdot R \qquad (2)$$

This means that R (EQ) also determines the velocity of ΔX and that a decrease (increase) in EQ means that foot tilt with a certain velocity will result in a slower (faster) CoP displacement and that faster (slower) foot tilt is necessary to generate CoP displacement with the same velocity.

According to the RSI, for small values of Δφ, the relation between ΔM and Δφ can be described as

$$\Delta M = h_M \cdot \Delta\varphi \qquad (3)$$

(appendix) with h_M as the height of the most lateral point of the lateral malleolus. With equations (1) and (3), EQ can be calculated by

$$EQ = R = \frac{\Delta X \cdot h_M}{\Delta M} \qquad (4)$$

When the magnitudes of both EQ and h_M are assumed to be constant, it can be inferred from equation (4) that ΔM and ΔX are linearly related. This means that we can evaluate the appropriateness of the RSI as a model for the foot by assessing the linearity of the relation between ΔM and ΔX.

Foot size

The maximum value of ΔX (ΔX_{max}) can be calculated as

$$\Delta X_{max} = \Delta\varphi_{max} \cdot R \qquad (5)$$

with $\Delta\varphi_{max}$ as the maximum value of $\Delta\varphi$. During one-leg stance the position of the CoP is always restricted to the area enveloped by the most peripheral points of application of force. This area is called the area of support. For the frontal plane this means that, irrespective of $\Delta\varphi_{max}$, ΔX_{max} is restricted to the breadth of the area of support. The magnitude of the latter is determined by the breadth of the foot. Consequently, it can be inferred from equation (5) that R and consequently EQ, are related to the breadth of the foot. More precisely, R and EQ are related to functional foot breadth, i.e. the breadth of the part of the foot supporting the body weight and being used for balance control. R and EQ do not represent functional foot breadth. According to the RSI, a change in foot loading towards the wide forefoot should result in an increase in the functional foot breadth and consequently in an increase in EQ. The opposite is true when foot loading is changed towards the narrow rearfoot.

When we assume that all feet have about the same shape, and that during erect stance the same part of the foot is being used for balance control, then, in erect stance, the functional foot breadth will be an approximately constant percentage of the maximum foot breadth (b_{f-max}). In that case the b_{f-max} of a subject can be taken as a measure for the functional foot breadth of that subject. When these assumptions are true, b_{f-max} should be linearly related with EQ. It can be concluded that, according to the RSI, both intra- and inter-subject variations in functional foot breadth will affect EQ.

Foot tilt strategy

In order to clarify the role of RSI mediated CoP displacement in balance control we first define the posture of equilibrium of the RSI-model. In the posture of equilibrium there is equilibrium of forces and moments of force about all possible centers of rotation. In the RSI-model, equilibrium depends on the force of gravity acting on the CoG (F_g), the ground reaction force (F_{rg}) and the distance between the working lines of these forces and the potential centers of rotation. Because $F_g = F_{rg}$ and because the moments about the subtalar joint and the CoP, caused by F_g and F_{rg} are zero in Figure (2A), this posture is defined as the posture of equilibrium. When the model deviates from this posture and starts to tilt, this movement can be regarded as a frontal plane rotation of the body about its center of rotation. When the body rotates about the supporting foot, the subtalar joint of the supporting foot can be regarded as the center of rotation (Figure 2B). However, when the muscles across the subtalar joint are contracted, the tilting movements of the body superior to the subtalar joint will be transmitted to the foot inferior to the subtalar joint. As a result of this the entire body rotates about a center that is located under the foot, probably in the vicinity of the CoP. In the RSI-model, this external center of rotation is located exactly in the center of pressure (Figure 2C).

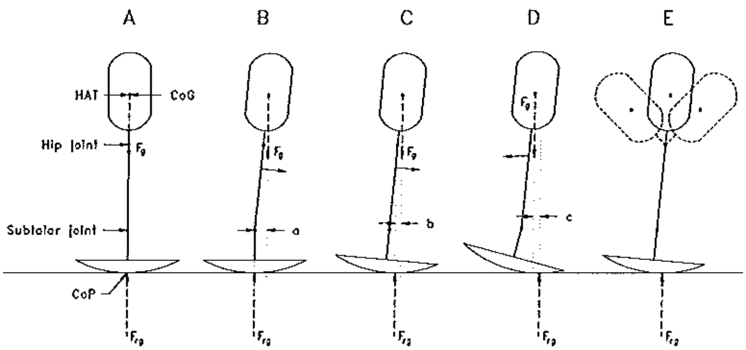

Figure 2. Equilibrium of forces ($F_{rg} = F_g$) and moments of force ($\Sigma M = 0$) (A), loss of equilibrium due to tilting of the body about the subtalar joint ($\Sigma M = F_g \cdot a$) (B) and tilting of the body about the CoP, ($\Sigma M = F_g \cdot b$) (C). Return towards the position of equilibrium by rotation about the subtalar joint and the CoP ($\Sigma M = -F_g \cdot c$) due to foot tilt (D). The absence of a continuous linking of the displacements of the head, arms and trunk (HAT) and the CoP (E).

To regard the CoP as the external center of rotation is not entirely without merit, because, when the foot is tilted from the neutral position to a position in which only the edge of the foot is in contact with the floor, it is conceivable that, like the CoP, the center of rotation displaces towards a position under the edge of the foot. Rotation about the subtalar joint or about the external center of rotation can be expected to occur alternately during one-leg stance as well as walking. This variability in the position of the center of rotation could be the cause of the modelling errors encountered when assuming that the CoG rotates only about the subtalar joint[18]. Rotation about the subtalar joint and about the CoP results in a destabilizing, gravitational moment of force ($F_g \cdot a$) and ($F_g \cdot b$) respectively (Figure 2B and C), which promotes further body tilt. Tilting the RSI counteracts this body tilt by displacing the CoP towards the opposite side of the line of action of F_g and generating a stabilizing moment of force ($-F_g \cdot c$) (Figure 2D). Due to this stabilizing moment of force the body returns towards the posture of equilibrium. We call this mechanism for balance control the Foot Tilt Strategy (FTS), because, according to the RSI, foot tilt is imperative for the displacement of the CoP. The FTS is an adaptation of the ankle strategy to the narrow support surface in frontal plane balance control during one-leg stance. It is less strictly on the function of the ankle joint as the body's sole center of rotation and puts more emphasis on foot tilt as source of CoP displacement. These differences manifest itself in the ways the supporting foot is modelled for the ankle strategy (Figure 1B) and the FTS (Figure 1D). It can be concluded that, according to the RSI model, the frontal plane movements of the CoP are the result of the activity of the FTS as a mechanism for balance control during one-leg stance.

MATERIALS AND METHODS

Subjects

We measured 6 male and 1 female subject with a mean age of 26 years (range 22-38 years), a mean height of 1.84 m (range 1.64-1.95), a mean weight of 75.5 kg (range 60.3-93.9) and a mean foot length and breadth of 265 mm (range 232-285) and 101 mm (range 92-112). None of the subjects had a history of orthopedic or neurological disorders. All subjects were students or university personnel.

Instruments

For stabilometry we used a forceplate with piezoelectric force transducers (Kistler type 9281B, Kistler Instrumente AG, Winterthur, Switzerland). The platform was connected to electronic amplifier units (Kistler types 5001, 5006 and 5675, Kistler Instrumente AG, Winterthur, Switzerland). ΔCoG is calculated by dividing F_x by the body mass followed by double integration.

The movements of the lateral malleolus of the left foot were measured with a displacement transducer (Schlumberger DF\5.0\S, Sangamo transducers, Bognor Regis, England) connected to the left one of two horizontally pivoting flaps between which the ankle is positioned (Figure 3). Because the two flaps of the ankle measuring apparatus were connected by an elastic band, the subject's lateral malleolus was permanently in contact with the flap connected to the displacement transducer. To prevent sagittal plane ankle movements from activating the displacement transducer, the supports on which the flaps were mounted were movable in the sagittal plane. To prevent displacement of the lateral malleolus relative to the left flap, a plastic block with an u-shaped groove, to accommodate the lateral malleolus, was mounted on the inside of the left flap (Figure 3).

We measured the left foot because the displacement transducer was mounted on the left side, and because the lateral malleolus is more protruding thus reducing the risk of displacement of the ankle relative to the flap connected to the displacement transducer.

The height of the most lateral part of the lateral malleolus (h_M) was measured with a ruler. $b_{f,max}$ was measured with a device in which infrared light emitting diodes and infrared sensors move parallel to each other along the length and breadth of the foot. The maximum length and breadth of the foot are measured with an accuracy of 1.0 mm. These devices are commercially available and are often used in shoe shops.

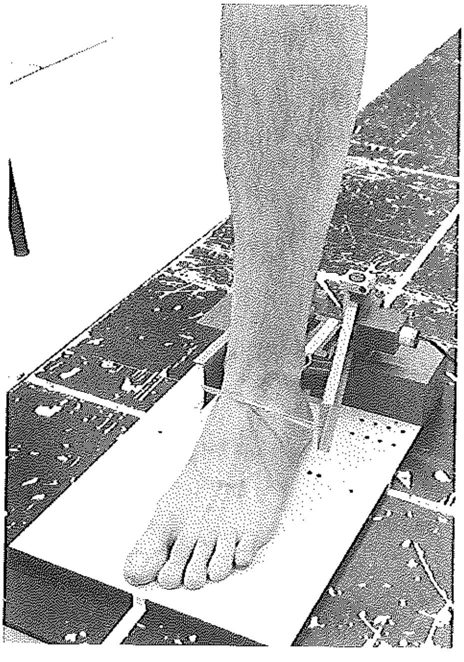

Figure 3. Displacement transducer operated device measuring the horizontal displacement of the lateral malleolus in the frontal plane.

Forceplate and displacement transducer signals were low-pass filtered with fourth order, Butterworth type filters (Kemo LTD, Beckenham, England; Krohn-Hite Corp., Avon, Mass., USA) with the cut-off frequency at 20 Hz. Output signals were recorded at 100 samples per second with a data acquisition board (DAS-1602, Keithly Metrabyte, Taunton, MA, USA) and fed into a personal computer for later analysis. Each measurement consisted of 1024 samples resulting in a test time of 10.2 sec.

To measure the movements of the upper body a reflective marker was attached to the back with adhesive tape. The marker was located in the midline at the level of the first thoracic vertebra. The movements of the marker were registered with a Sony AVC-D5CE CCD camera (Sony, Tokyo, Japan) connected to a 68030 processor based VME computer system[22]. Video signals were recorded with 50 Hz and digitally filtered twice with a second order Butterworth type filter[17] with a final cut-off frequency of 5 Hz.

Procedure

After they had been briefed about the purpose and the procedure of the experiment, the subjects gave written informed consent and entered the experiment. Measurements took place in a well-lighted room. Subjects were asked to use a black dot 24 mm in diameter, attached to the wall 3.6 m in front of them, as a visual reference point and stand as motionlessly as possible. To reduce the effect of intra subject variability, each subject was measured three times; the mean value of these three repetitions was calculated. The measurement position was: left foot in the center of the forceplate, long forceplate axis through first interdigital space and middle of the heel, short forceplate axis through medial malleolus. The supporting knee was straight but not hyperextended. To prevent their use in balance control, the non-supporting knee was bent approximately 90° and held against the contralateral knee and both arms were crossed and held against the chest. The reason for this was to let the subject and the linked segment model in Figure (2A) be as congruous as possible with respect to the available mechanisms for balance control. After having assumed the correct measurement posture the subjects were allowed 10 sec to settle before the start of data acquisition. Between consecutive measurements the subjects had a 1-minute break to prevent fatigue.

Statistical analysis

The appropriateness of the assumption of a linear relationship between ΔX and ΔM was assessed by examining all the 1024 data points in each simultaneous measurement of ΔX and ΔM with linear regression with ΔX as the dependent variable. The linearity of the relation between EQ and $b_{f\text{-}max}$ was examined with linear regression with EQ as the dependent variable. The relation between ΔX on the one hand and ΔCoG, ΔF_x, ΔU, ΔF_y and ΔY on the other hand was examined with the same method, again with ΔX as the dependent variable. Besides the relation between ΔX and ΔY we examined the linearity of the relation between ΔF_x and ΔF_y to assess the amount of crosstalk between the frontal and the sagittal plane. We calculated the correlation coefficient (r) which describes the strength of the association between the two variables and the coefficient of determination (r^2), describing how clearly a straight line describes the relationship between these two variables[23]. The obtained values for r and r^2 are only valid within the assessed range of motion. The median value of ΔX is used as a measure for the latter. The value of β is the slope of the regression line and was used for the quotient ΔX/ΔM in the calculation of EQ.

As an additional measure of linearity and to determine to what extent ΔX and the other measured variables were interchangeable we calculated the standard deviation (s) of the difference between the value of ΔX predicted by the linear model (ΔX_p) and the actually measured value of ΔX. Assuming that these differences followed a Normal (Gaussian) distribution, 95% of the actual values will lie in the interval between $\Delta X_p + 1.96s$ and $\Delta X_p - 1.96s$ of the regression line[24]. The width of this prediction interval is therefore calculated as

$2 \cdot 1.96 \cdot s$. It depends on the relationship between the width of the prediction interval and the magnitude of the effects on CoP displacement encountered in stabilometry whether these variables can be used interchangeably. All statistics were calculated with the statistical program SPSS/PC+™ (version 5.02); other calculations were done with the program Quattro Pro™.

RESULTS

Compared to the other measured variables ΔM showed the strongest correlation and consequently the most linear relation with ΔX (Table 1). The value of s for the relation between ΔX and ΔM (Table 1) resulted in a 9.4 mm wide prediction interval. The average values for ΔX and ΔM were 5.07 (SD 0.81) and 1.25 (SD 0.46) mm respectively. Although the relation between ΔX on the one hand and F_x and ΔU on the other was about 40% less linear than the relation between ΔX and ΔM, it still showed a considerable amount of linearity (Table 1).

Variable	r	r^2	s	β
ΔM	.91 (SD 0.04)	.83 (SD 0.07)	2.41 (SD 0.43)	$4.18^{(0)}$ (SD 1.00)
ΔCoG	.49 (SD 0.18)	.32 (SD 0.17)	4.88 (SD 0.99)	$0.77^{(3)}$ (SD 0.40)
ΔF_x	.65 (SD 0.11)	.45 (SD 0.13)	4.67 (SD 1.23)	$1.16^{(0)}$ (SD 0.32)
ΔU	.73 (SD 0.06)	.54 (SD 0.08)	4.08 (SD 0.59)	$0.59^{(0)}$ (SD 0.22)
ΔF_y	.23 (SD 0.10)	.07 (SD 0.04)	5.97 (SD 0.88)	$-0.21^{(3)}$ (SD 0.75)
ΔY	.21 (SD 0.07)	.06 (SD 0.05)	6.01 (SD 0.98)	$-0.14^{(3)}$ (SD 0.21)

Table 1. Mean values of the correlation coefficient (r), the coefficient of determination (r^2), the standard deviation of the difference between ΔX and ΔX_p (s) and the slope of the regression line (β) between ΔX and the variables in column 1. In superscript the number of non-significant values of β on a total of 21. (n = 7)

The correlation between ΔX and ΔCoG on the other hand was low, especially in view of the formerly assumed relation between these variables. The linearity of the relations between ΔX on the one hand and F_y and ΔY on the other were the lowest encountered here (Table 1). The relation between ΔF_x and ΔF_y was characterized by values for r, r^2 and s of .21, .05 and 3.65, respectively. When we took a closer look at ΔX as a function of ΔM we observed that it followed an ellipsoid course in a clockwise direction (Figure 4). Each measurement consists of multiple ellipsoids in line, of which only three are shown in figure 4.

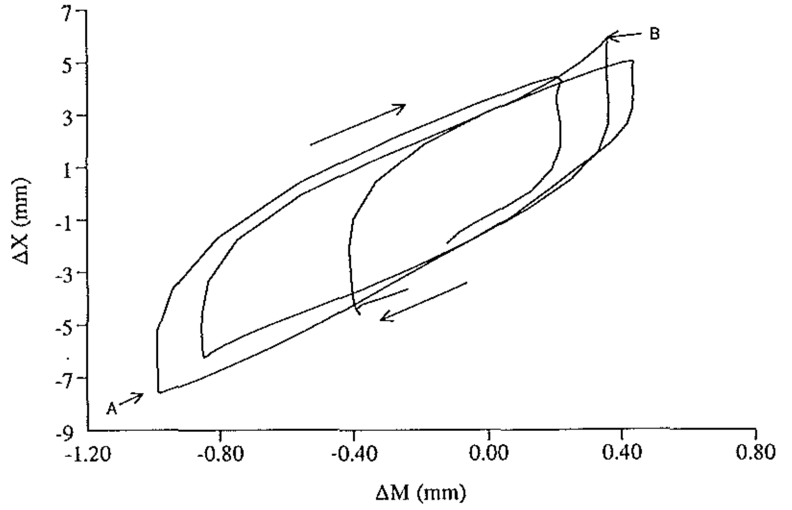

Figure 4. A clear example of the clockwise, ellipsoid course of the CoP (ΔX), as a function of the horizontal displacement of the lateral malleolus of the left foot, in the frontal plane (ΔM). The arrows indicate the medial (A) and lateral (B) turning points where the strongest deviation from linearity occurs. This could be the result of forefoot movements that are independent of ankle displacement.

Regression analysis of the subject specific values of EQ and $b_{f\text{-}max}$ (Figure 5) resulted in values for r, r^2, s and β of .85 .72, 66 and 15.7 (p=.016), respectively. Without the subject with an EQ of 523 the values for r, r^2, s and β were .96, .92, 31 and 13.9 (p=.003), respectively.

DISCUSSION

The model

The values of r and r^2 (Table 1) show that the relation between ΔX and ΔM is predominantly linear. This means that the RSI offers a fairly good description of the relation between ΔX and ΔM during one-leg stance. Theoretically, according to the RSI, the correlation coefficient between ankle movement and CoP movement should be 1. However, there are several possible causes for the difference between the theoretically predicted and the experimentally derived correlation coefficient. First, the assumption that EQ is fixed, which was adopted after the introduction of equation (4), is probably incorrect because during one-leg stance the distribution of the body weight over forefoot and rearfoot changes continuously.

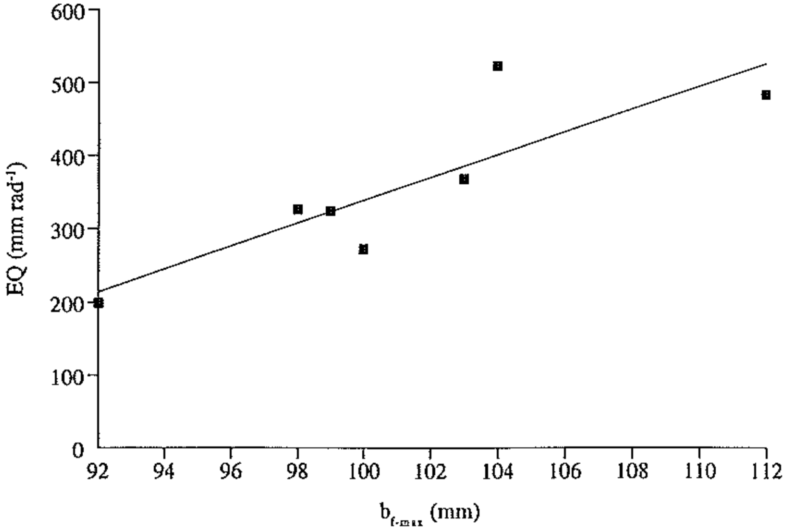

Figure 5. The efficiency quotient of foot tilt (EQ) as a function of maximum foot breadth ($b_{f\text{-}max}$) with the matching regression line. The outlier with an EQ value of 523 belonged to a subject with a wider midfoot due to flatfoot deformity.

According to the RSI this will cause variations in EQ and consequently a decrease in linearity. Second, the relation between ΔM and $\Delta\varphi$ as proposed in equation (3) is an approximation of the actual relation between these variables. Third, because the foot is composed of multiple bones the forefoot can move relative to the rearfoot and forefoot movements can cause CoP displacement independent of ankle displacement. Nevertheless, because of the high values of r and r^2 for the relation between ΔX and ΔM, both absolute as well as relative to the values of r and r^2 for the relations between ΔX and the other recorded variables, we accepted the RSI as a suitable basis for further exploration of the relation between ankle and CoP displacement. The average values for ΔX found in this study is comparable with the values during one-leg stance found elsewhere[7, 25]. Consequently, the validity of this model clearly encompasses the ranges of motion usually encountered during one-leg stance.

To decide whether measurement of ΔM and ΔX can be used interchangeably, we compare the width of the prediction interval (9.4 mm) with the differences in CoP displacement often encountered in stabilometry. Because the latter are in the order of a few mm, we concluded that the interchangeability of ΔM and ΔX is limited and that these variables can only be used as an approximation of each other.

The relatively high correlation between ΔX and ΔF_x (Table 1) can be explained by the fact that, besides moving the CoP, foot tilt implies movement of a part of the body and therefore displacement of the CoG and generation of F_x, and because the foot is in close contact with the forceplate its movements will be transmitted to the forceplate undampened. During undisturbed one-leg stance FTS activity may be the most prominent frontal plane body movement and consequently the major source of F_x.

The high correlation of ΔX with ΔU seems surprising considering the distance and the presence of at least two joints between the foot and the upper body. However, it follows from the aforementioned description of FTS activity during total body balance control (Figure 2A-D) that ΔX and ΔU partly move in the same direction. This can explain the high correlation between these variables.

The mean correlation between ΔX and ΔCoG is surprisingly low considering the fact that CoP displacement is often interpreted as a measure of body sway. However, the large SD for the relation between these variables (Table 1) shows that considerable inter-subject differences occurred. In fact, there were also large intra-subject differences. It was our impression that more upper body movements, i.e. increased hip strategy, resulted in a lower correlation between ΔX and ΔCoG. This situation is sketched in Figure (2E). In view of the fact that the HAT segments comprises about 60% of the bodyweight there seems to be a discrepancy between the correlation between ΔX and ΔU and between ΔX and ΔCoG. This is probably caused by the position of the video marker. Due to the location of this marker at the level of the first thoracic vertebra instead of the CoG of the HAT segment, hip strategy activity can cause a discrepancy between the displacement of U and the CoG. In the absence of hip strategy activity the body moves more like an inverted pendulum and ΔU and ΔCoG will both correlate strongly with ΔX. This would also explain the high SD value for the correlation between ΔX and ΔCoG. More research on this subject seems necessary.

The low correlations between ΔX and ΔY and between ΔF_x and ΔF_y, indicate that, for these variables, there is only a minor amount of crosstalk between frontal- and sagittal plane. Consequently, effects on frontal- or sagittal-plane balance control can be regarded as the result of events occurring in these respective plains and not as the result of crosstalk from events occurring in a perpendicular plane.

When we compare the correlations found between ΔX and other variables with the corresponding values found by Goldie et al.[26] we must bear in mind that these authors calculated correlations between the standard deviations of various forceplate parameters and that we calculated the correlations between the variables themselves. A possible source of discord is the fact that the standard deviations of two variables representing entirely different aspects of the movements of the body, can still be strongly correlated because both variables are somehow affected by the amount of body movements. The opposite, a strong correlation between the variables themselves and a low correlation between their standard deviations, is not possible. This is corroborated by the correlations (this paper v. Goldie) between ΔX on the one hand and ΔY (.21 v. .39), ΔF_y (.23 v. .77) and ΔF_x (.65 v. .82) on the other. The correlations found in this study are all lower.

It can be inferred from the ellipsoid course of ΔX as a function of ΔM (Figure 4) that the strongest deviation of linearity is concentrated at the turning points, where CoP displacement without ankle movement takes place. The opposite, i.e. ankle movement without CoP displacement, does not occur. A possible explanation for this is the earlier mentioned forefoot flexibility that allows forefoot movements independent of foot tilt. According to this explanation a single ellipse could be interpreted as follows: starting from point A (medial) the CoP moves towards lateral due to the contraction (relaxation) of muscles acting on the forefoot independent of foot tilt. Next ΔM and ΔX displace predominantly parallel towards lateral until point B. From point B (lateral) ΔX initially moves to medial due to the relaxation (contraction) of the aforementioned muscles, followed by a predominantly parallel displacement of ΔX and ΔM to medial. When this interpretation is correct, the width of the ellipse could be used as a measure for the contribution of forefoot movements to balance control. Although none of the other factors mentioned as potential sources of non-linearity is expected to operate predominantly at the turning point of the ankle movements, the above explanation is hypothetical and more research on this subject is necessary.

Balance control

Because the frontal plane movements of the CoP are predominantly the result of FTS activity, an increase in the amplitude or velocity of the displacement of the CoP could be interpreted as an increase in the activity of this mechanism for balance control. This can occur in situations where balance control is more arduous, e.g. with increasing age or after eye closure[27], ankle[7] or knee[9] injuries . A decrease in the amplitude or velocity of the displacement of the CoP could be interpreted as decreased FTS activity. The latter can occur because balance control is less arduous, e.g. due to training[8, 16] or because foot tilt is constrained. However, in view of the role of EQ in the relation between ΔM and ΔX, one must always consider the possibility of a change in EQ when interpreting a change in ΔX as a change in FTS activity. The importance of the FTS for balance control is illustrated by the fact that people with a subtalar[28] or a talonavicular[29] arthrodesis, i.e. with limited foot tilt, experience balancing problems when walking on uneven ground.

To contribute to balance control, a certain amount of CoP displacement must be attained within a certain amount of time. This means that demands are being made on the amplitude and velocity of CoP displacement. During one-leg stance, the amplitude and the velocity of CoP displacement depend on the amplitude and the velocity of foot tilt as well as on EQ. This means that, for foot tilt with a given amplitude and velocity, a decrease in EQ will cause a decrease in the amplitude and velocity of CoP displacement. Unless this decrease in EQ can be adequately compensated for by an increase in the amplitude and velocity of foot tilt it will result in a decrease in the contribution of the FTS to balance control which, in turn, could have a negative effect on balance control. However, even when a decrease in EQ is fully compensated by an increase in the amplitude and velocity of foot tilt, a negative effect on balance control is still possible because the spatial accuracy of rapid aimed movements is known to be inversely proportional to the velocity of the movements[30, 31]. This means that foot tilt, and consequently CoP displacement, would be less coordinated. This means that any decrease in the value of EQ, whether the result of internal (anthropometric) or external (footwear) factors, could have a negative effect on balance control.

Unlike the horizontal, frontal plane component of the reaction force vector (F_x), which is affected by FTS activity as well as upper body movements, CoP displacement predominantly reflects the activity of the FTS. This interpretation of CoP displacement is much more restricted than the one in which CoP displacement is thought to represent body sway. This more restricted interpretation of CoP displacement explains the lower sensitivity and predictive value regarding steadiness of CoP based variables compared to F_x based variables as found by Goldie et al.[26, 32].

Foot breadth

The hypothesized linearity between EQ and $b_{f\text{-}max}$ is confirmed by the high values of r and r^2 and the positive and significant value for β, especially after removal of the outlier with an EQ of 523 (Figure 5). Unlike the others, this subject appeared to have a flatfoot deformity with widening of the midfoot due to flattening of the medial, longitudinal arc. This probably caused the high value of EQ. Both findings corroborate the hypothesis that differences in foot breadth are associated with differences in EQ. Theoretically, an artificially induced increase in foot breadth e.g. due to footwear or an orthosis, should also cause an increase in EQ and possibly improve balance control. This is corroborated by the fact that shoes improve balance when walking on a beam[33, 34]. Further investigation of the effect of foot size and footwear on balance control seems warranted.

Conclusions

In this study a model for the relation between horizontal, frontal plane ankle displacement and CoP displacement during one-leg stance is described and validated. According to this model the frontal plane displacement of the CoP does not represent body sway, but is the result of foot tilt as a mechanism for balance control during one-leg stance. It was shown that foot breadth affects the gain of foot tilt in terms of CoP displacement. Because CoP displacement plays an important role in balance control, foot breadth as well as other factors altering this gain can be expected to affect balance control.

APPENDIX

ΔM due to tilting of the RSI has a translatory (ΔM_t) as well as a rotatory (ΔM_r) component.

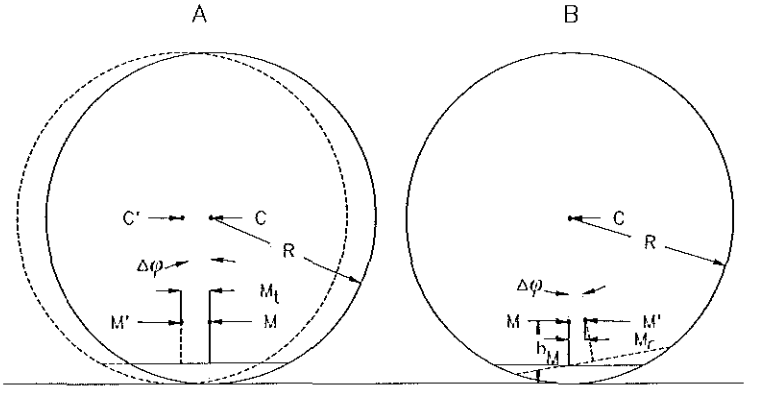

Figure 6. The counteracting translatory (ΔM_t) (A) and rotatory (ΔM_r) (B) components of the displacement of the lateral malleolus (M) are a function of: the amount of (counterclockwise) tilt of the RSI ($\Delta\varphi$), the height of M (h_M) and the radius of the circle with center C, of which the curved surface of the RSI is a part.

When the RSI tilts to the left and the angular displacement is $\Delta\varphi$, ΔM_t (Figure 6A) and ΔM_r (Figure 6B) are,

$$\Delta M_t = R \cdot \Delta\varphi \tag{A1}$$

and

$$\Delta M_r = h_M \cdot \sin\Delta\varphi - R \cdot \sin\Delta\varphi \tag{A2}$$

respectively.

According to equations (A1) and (A2) the relation between ΔM and Δφ can be described as

$$\Delta M = h_{M_f} \cdot \sin\Delta\varphi + R \cdot \Delta\varphi - R \cdot \sin\Delta\varphi \tag{A3}$$

For small angles Δφ this can be simplified to

$$\Delta M = h_{M_f} \cdot \Delta\varphi \tag{A4}$$

Theoretically we should measure ΔM at the level of the subtalar joint. However, we actually measured ΔM at the most lateral point of the lateral malleolus which is about 12 mm more superior[35] (Figure 7).

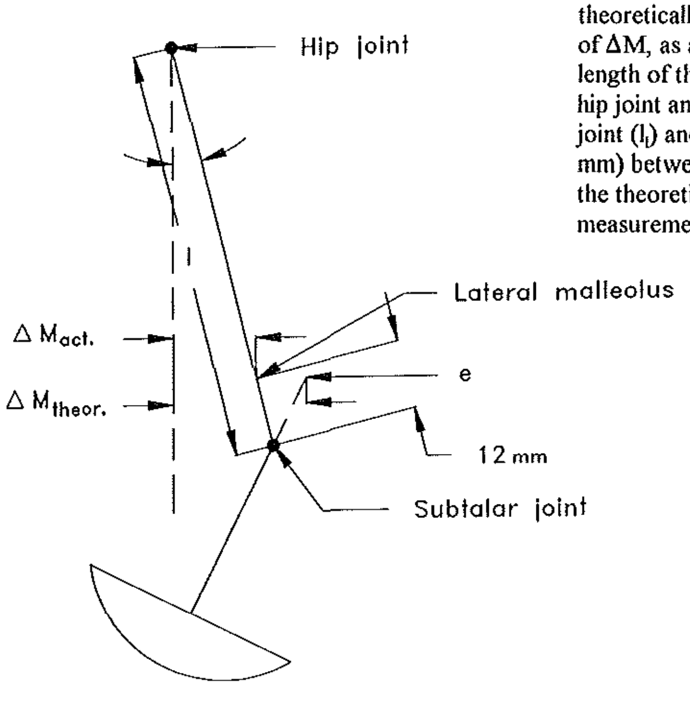

Figure 7. The discrepancy between the actually and the theoretically measured value of ΔM, as a function of the length of the leg between the hip joint and the subtalar joint (l_l) and the distance (12 mm) between the actual and the theoretical (e) measurement position of M.

However, when we measure a subject with a leg length l_l (Figure 7), the error resulting from the 12 mm discrepancy between the theoretical and the actual position of the point of registration of ΔM in terms of percentage is calculated as

$$Error\ \ \Delta M = 100 \cdot \frac{2 \cdot 12 \cdot \sin(\alpha)}{l_l \cdot \sin(\alpha) + 12 \cdot \sin(\alpha)} \tag{A5}$$

with α as the angle of the segment between the hip joint and the subtalar joint, relative to the working line of the force of gravity. In a subject with a body height of 180 cm this will result in a leg length of 884 mm[36], and an error in ΔM of 2.7%. Because this error is quite small, equation A4 can be used to describe the relation between Δφ and ΔM.

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Chapter III

Effect of increased compliance of the supporting surface on frontal plane balance control, during one-leg stance

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Submitted

List of abbreviations

CoP	center of pressure
CoG	center of gravity
EQ	the efficiency quotient of foot tilt
FTS	foot tilt strategy, a mechanism for balance control
HAT	body segment comprising of head, arms and trunk
RSI	rocker shaped interface with the supporting surface, representing the supporting foot
φ	angle of frontal plane foot tilt
h_M	height of the lateral malleolus
K	the spring constant or stiffness of the springs, reflecting foam compliance
M	the horizontal, frontal plane position of the lateral malleolus
R	the radius (mm) of the circle of which the curved surface of the RSI is a part
SB	suspended bar, representing the foot and its supporting surface
U	the horizontal, frontal plane position of the upper body
X	the frontal plane position of the CoP relative to the forceplate coordinate system
Y	the sagittal plane position of the CoP relative to the forceplate coordinate system
F_g	the force of gravity acting on the body's CoG
F_{rg}	the vertical ground reaction force acting on the supporting foot
F_n	the vertical ground reaction force under the left (n=1), right (n=2) spring, left foot (n=l) or right foot (n=r)
F_x	frontal plane component of the horizontal ground reaction force
b	halve width of the foot
A_X	the median amplitude of the displacement of the CoP in the frontal plane
A_Y	the median amplitude of the displacement of the CoP in the sagittal plane
A_M	the median amplitude of the displacement of the lateral malleolus in the frontal plane
v_X	the median amplitude of the velocity of the CoP in the frontal plane
v_Y	the median amplitude of the velocity of the CoP in the sagittal plane
v_M	the median amplitude of the velocity of the lateral malleolus in the frontal plane

INTRODUCTION

Balance control depends on the input of the visual, vestibular and proprioceptive (cutaneous, joint and musculotendinous) system and the output of the musculoskeletal system. A much practiced method to assess balance control is platform stabilometry: here subjects stand on a forceplate that measures the vertical, fore-aft and medio-lateral components of the ground reaction force vector. From these forces, several parameters related to the movements of the subject standing on the forceplate can be calculated.

To test the system governing balance control, the balance control task is often made more difficult. One method to make balance control more arduous is to place subjects on a sheet of foam (Bles and De Wit 1975; Brandt et al. 1981; Di Fabio and Badke 1991; Ledin et al. 1993; Norre 1992 and 1993; Teasdale et al. 1991; Weber and Cass 1993; Yardley et al. 1992). In all these studies the implicit assumption is made that the effect of foam on balance control occurs solely via a decrease in proprioception. It is debatable, however, whether or not this is correct.

Some of the most popular parameters in platform stabilometry are based on the movements of the Center of Pressure (CoP). The CoP is defined as the point of application of the resultant vertical reaction force vector (F_{rg}). To contribute to balance control, the CoP must be continuously moving ahead and behind the body's center of gravity (CoG; Winter 1990). This means that demands are made on the coordination, the amplitude and the velocity of CoP displacement. During platform stabilometry, CoP displacement depends on changes in the pressure distribution under the supporting foot (feet). It can be inferred that changes in the compliance of the supporting surface affect the interaction between the foot (feet) and this surface. This is expected to result in a decrease in the efficiency with which CoP displacement is generated, in such a way that balance control can become deteriorated.

During one-leg stance, frontal plane CoP displacement is predominantly the result of frontal plane foot tilt as a mechanism for balance control (Hoogvliet et al. in press 1). This mechanism was called the foot tilt strategy (FTS). To reflect the linear relation between frontal plane foot tilt and frontal plane CoP displacement, the supporting foot is modeled as a rocker-shaped interface (RSI; Figure 1A) with the supporting surface. RSI tilt is consistently linked with the movements of the CoP because the point of contact between the RSI and the supporting surface is their only point of contact and therefore equals, at any time, the position of the CoP (Figure 1B). The relation between foot tilt and CoP displacement is characterized by EQ, defined as the amount of CoP displacement per radian of foot tilt. EQ is numerically equal to the radius (R) in mm, of the circle of which the curved surface of the RSI is a part (Figure 1A). During one-leg stance the amplitude and the velocity of CoP displacement depend on the amplitude and the velocity of foot tilt, and on EQ. This means that a decrease in EQ will cause a decrease in the amplitude and velocity of CoP displacement unless there is a compensatory increase in the amplitude and velocity of foot tilt. Unless this compensation restores the amplitude and velocity of CoP displacement to their previous levels, there will be a decrease in the contribution of the FTS to balance control. This, in turn, will result in a deterioration of balance control. We hypothesize that standing on foam decreases the value of EQ.

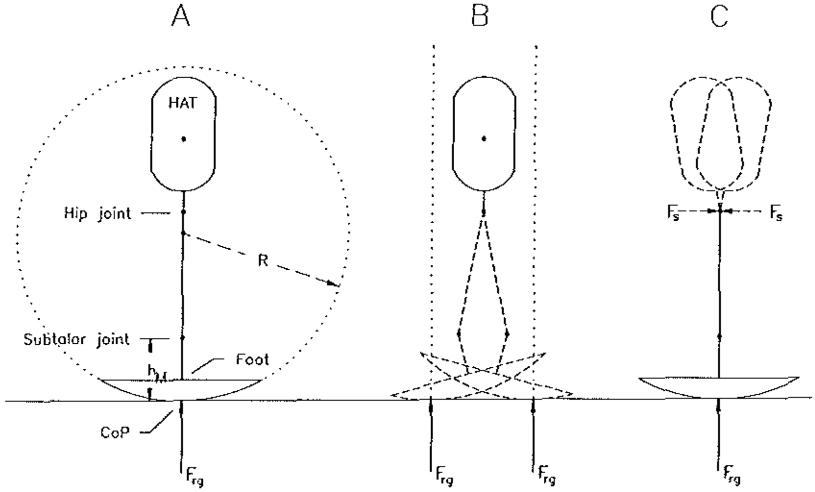

Figure 1. A linked segment model of the human body during one-leg stance (A). Head, arms and trunk are modeled as the HAT segment (A). The foot is modeled as a rocker-shaped interface (RSI) with the supporting surface. RSI tilt is consistently linked with the movements of the CoP because the point of contact between the RSI and the supporting surface is their only point of contact and therefore equals, at any time, the position of the CoP (B).

The purpose of this study is to establish an alternative effect of foam on balance control, i.e., besides decreased proprioception. We expected the altered interaction between the foot (feet) and the supporting surface to result in impeded CoP displacement. First the interaction of the foot (feet) and its supporting surface was modeled. With this model we substantiated the hypothesized effect on impeded CoP displacement for one and two-leg stance. This hypothesis was tested for one specific situation, i.e., frontal plane foot tilt during one-leg stance, by assessing the effect of standing on foam on EQ. Finally, we assessed the effect of altered EQ on balance control.

METHODS

Modeling

We modeled a two-dimensional foot (Figure 2A) on a solid surface as a bar (the foot) suspended by springs with infinite stiffness (the solid surface).

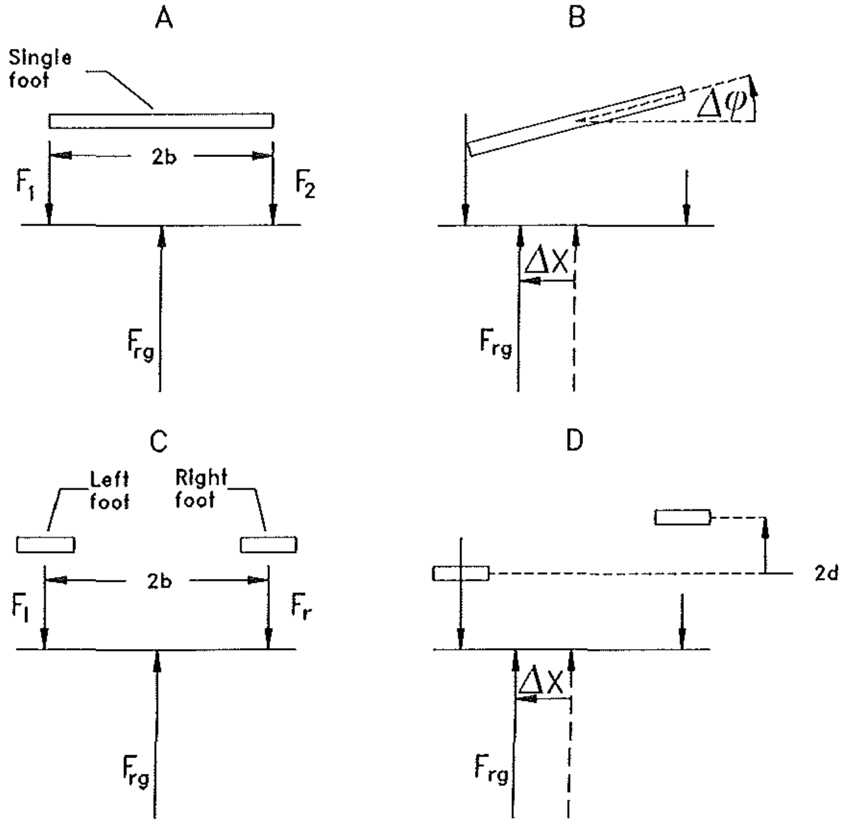

Figure 2. Displacement of the CoP, i.e., the point of application of F_{rg} in one (A-B) and two-leg stance (C-D). During one-leg stance, CoP displacement (ΔX) is the result of changes in F_1 and F_2 due to tilting (angle $\Delta\varphi$) of the foot (horizontal bar). During two-leg stance, ΔX is the result of changes in F_l and F_r due to lifting (distance 2d) of a foot (horizontal bar). For abbreviations, see text.

Similar to the geometric RSI model, this model couples foot tilt and CoP displacement. However, the suspended bar (SB) model is more realistic than the RSI because, as in the real foot, CoP displacement in the SB model, is the result of alterations in the pressure distribution under the foot. In the SB model, the point of application of F_{rg} depends on the relation between the left (F_1) and right (F_2) spring to ground force (Figure 2A). The relation between these three forces is given by

$$F_1 + F_2 = F_{rg} \tag{1}$$

Tilting of the foot alters F_1 and F_2 and moves F_{rg} over a distance ΔX (Figure 2B). The efficiency with which the CoP is displaced can be calculated as follows. The moments equation about the point of application of F_2 can be described as

$$F_{rg} \cdot (b + \Delta X) = F_1 \cdot 2b \tag{2}$$

with 2b as the foot width and ΔX as the amount of CoP displacement due to foot tilt. When the foot is tilted over an angle $\Delta\varphi$ (rad), the equilibrium of forces yields

$$F_1 = \frac{1}{2} F_{rg} + K \cdot b \cdot \sin(\Delta\varphi) \tag{3}$$

and

$$F_2 = \frac{1}{2} F_{rg} - K \cdot b \cdot \sin(\Delta\varphi) \tag{4}$$

with K as the spring constant or stiffness of the springs, reflecting foam compliance. For small angles of $\Delta\varphi$, equation (2) and (3) can be rewritten as

$$\frac{X}{\Delta\varphi} = EQ = \frac{2 \cdot K \cdot b^2}{F_{rg}} \tag{5}$$

Equation (5) shows that, like the RSI model, the SB model describes the relation between foot tilt and CoP displacement as linear. Equation (5) also confirms the assumption that standing on foam, due to a decrease in K, results in a decrease in the efficiency with which foot movements move the CoP. The softer the foam, the smaller K and the lower the efficiency. It also states that EQ increases in parallel with foot breadth. This is in agreement with the positive relation between maximal foot breadth and the average value of EQ (Hoogvliet et al. in press 1). However, this relation was more a linear than a square one. This difference can be explained by the fact that the effects of b and F_{rg} on EQ are opposite, and that heavier subjects generally have larger feet. Consequently, the effect of an increase in b will be counteracted by an increase in F_{rg}.

It seems clear that the effect of foam is not restricted to the one-leg stance situation. To substantiate this, a SB model for two-leg stance was formulated. During two-leg stance the CoP can be displaced by foot tilt as well as foot lift. In the case of foot tilt, the effect of foam on each foot can be described with equation (5), after halving F_{rg}. During two-leg stance, CoP displacement due to foot lift depends on changes in the spring to ground force under the left (F_l) and the right (F_r) foot (Figure 2C-D). Consequently, equation (5) has to be rewritten as

$$EQ_{footlift} = \frac{\Delta X}{d} = \frac{2 \cdot K \cdot b}{F_{rg}} \qquad (6)$$

$EQ_{footlift}$ being the amount of Cop displacement per mm of foot lift. Two times b represents the horizontal distance between the feet and 2d is the vertical distance between the feet, representing foot lift. Equation (5) shows that $EQ_{footlift}$ is also decreased by foam. It is also shown that increasing the horizontal distance between the feet will increase the efficiency with which foot lift moves the CoP and facilitates balance control. This is corroborated by the fact that increasing stance width improves balance control (Kirby et al. 1987).

Variables

EQ is calculated with the following, simplified, equation:

$$EQ = \frac{\Delta X \cdot h_M}{\Delta M} \qquad (7)$$

the derivation of which was described earlier (Hoogvliet et al. in press 1).

In this formula, h_M is the height of the lateral malleolus in mm at the level of its most lateral point, and ΔX and ΔM are the horizontal, frontal plane displacement in mm of the CoP and the lateral malleolus, respectively. The average values of ΔX and ΔM in a single measurement are the median amplitude of the displacement of the CoP (A_X) and the lateral malleolus (A_M) in the frontal plane. A_M and the median amplitude of the velocity of the lateral malleolus (v_M) in mm and mm s^{-1}, respectively, represent the amplitude and velocity of foot tilt. The frontal plane movements of the CoP are characterized by A_X and the median velocity of the CoP (v_X) in mm and mm s^{-1}, respectively.

Frontal plane balance control is often assessed by measuring CoP displacement (A_X) or the horizontal ground reaction forces (F_x). The smaller these parameters are, the less the body moves and the better balance control is judged to be. Unlike A_X, which is predominantly the result of FTS activity, F_x represents the accelerations of the body's center of mass and takes into account both FTS activity and upper body movements. This explains the lower sensitivity and predictive value regarding steadiness of CoP based variables compared to F_x based variables, as found by Goldie et al. (1989 and 1992). Another argument in favor of F_x as the more appropriate variable to represent balance control is that A_X no longer consistently represents foot tilt when EQ changes.

When, during one-leg stance, foot loading is altered towards the wide forefoot or the narrow rearfoot this results in an increase and a decrease in EQ, respectively (Hoogvliet et al. in press 2). Changing foot loading towards anterior or posterior will also result in a respective increase and decrease in the position of the CoP in the sagittal plane (Y). To detect the presence of such an effect on EQ we calculated the mean value of Y in mm, relative to the forceplate coordinate system.

An increase in the compliance of the supporting surface is also likely to affect sagittal plane balance control. Although altered sagittal plane variables cannot be explained in terms of altered FTS activity, they do give information about sagittal plane balance control. To get an impression of the effect of standing on foam on sagittal plane balance control, the following sagittal plane variables are mentioned here. A_Y is the sagittal plane amplitude of the displacement of the CoP in mm. v_Y is the sagittal plane velocity of the CoP in mm s^{-1}. F_y is the sagittal plane component of the horizontal ground reaction forces (N), representing the horizontal component of the acceleration of the body's center of mass in that plane.

Subjects

Seven male and one female subject were measured: mean age was 24 (range 22-30) years, mean height was 1.84 (range 1.68-1.95) m, mean weight was 76.8 (range 62.3-94.4) kg, and all were without a history of major orthopedic or neurological disorders. All subjects were students or university personnel.

Prior to participating in the experiment all subjects gave written informed consent. This study has been performed in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki.

Instruments

For platform stabilometry we used a forceplate with piezoelectric force transducers (Kistler type 9281B, Kistler Instrumente AG, Winterthur, Switzerland). The platform is connected to electronic amplifier units (Kistler types 5001, 5006 and 5675). The horizontal, frontal plane movements of the lateral malleolus of the left foot were measured with a displacement transducer (Schlumberger DF\5.0\S, Sangamo transducers, Bognor Regis, England). The ankle measuring device incorporating this displacement transducer was described earlier (Hoogvliet et al. in press 1). Forceplate and displacement transducer signals were low-pass filtered with Butterworth type filters (Kemo LTD, Beckenham, England; Krohn-Hite Corp., Avon, Mass., USA) with the cut-off frequency at 20 Hz. Output signals were recorded at 100 samples per second with a data acquisition board (DAS-1602, Keithly Metrabyte, Taunton, MA, USA) and fed into a personal computer for later analysis. Each measurement consisted of 1024 samples resulting in a test time of 10.2 s. The compliant surface consisted of a 5 cm thick sheet of foam with a specific weight of 400 N m^{-3} (40 g/dm^3).

Procedure

Measurements took place in a well-lighted room. Subjects were asked to use a black dot 24 mm in diameter, attached to the wall 3.6 m in front of them, as a visual reference point and stand as motionlessly as possible. To reduce the intra-subject variability we averaged the outcome of (three) consecutive measurements. The positioning of the subject on the forceplate was described earlier (Hoogvliet et al. in press 1). After having assumed the correct measurement position the subjects were allowed 10 s to settle before the start of data acquisition. Between consecutive measurements the subjects had a 30 s break to prevent fatigue.

Statistical analysis

To assess the effect of increased compliance of the supporting surface, t-tests for paired samples were conducted on the averaged values of EQ, A_M, v_M, A_X, v_X, Y, A_Y, v_Y, F_x and F_y. A p-value $\leq .05$ was considered statistically significant. Statistical analysis was performed with the program SPSS/PC+™ (version 5.02).

RESULTS

Standing on foam resulted in a 40% decrease in EQ (Table 1). This decrease in EQ was paralleled by an increase in A_M and v_M (Table 1). The final result of these changes was that neither A_X nor v_X were altered significantly. Although standing on foam resulted in an increase in F_x the effect was not significant. Standing on foam resulted in a non-significant increase of Y. All other sagittal plane parameters also increased: with the exception of v_Y and F_y, these changes were not significant (Table 1).

Effect of a foam surface on balance control

Variable	Solid surface	Foam surface	p-value
EQ (mm/rad)	432 (SD 162)	260 (SD 56)	.008
A_M (mm)	1.00 (SD 0.51)	1.48 (SD 0.38)	.010
v_M (mm/s)	3.20 (SD 2.46)	5.43 (SD 2.91)	.012
A_X (mm)	4.69 (SD 1.17)	4.70 (SD 0.98)	.995
v_X (mm/s)	26.72 (SD 7.96)	25.64 (SD 7.08)	.613
Y (mm)	64.80 (SD 14.37)	66.01 (SD 9.75)	.625
A_Y (mm)	6.00 (SD 0.91)	5.39 (SD 1.33)	.291
v_Y (mm/s)	22.29 (SD 4.44)	24.02 (SD 5.22)	.028
F_x (N)	2.36 (SD 0.95)	2.54 (SD 1.28)	.371
F_y (N)	1.59 (SD 0.41)	1.73 (SD 0.45)	.029

Table 1. The effect of a 5 cm thick sheet of foam under the supporting foot during one-leg stance on: the amount of CoP displacement per radian of foot tilt (EQ), the amplitude and velocity of the horizontal displacement of the lateral malleolus (A_M and v_M), the amplitude and velocity of the displacement of the CoP in the frontal (A_X and v_X) and sagittal (A_Y and v_Y) plane, the horizontal component of the reaction force in the frontal (F_x) and sagittal (F_y) plane, and the mean position of the CoP in the saggital plane (Y). n = 8

DISCUSSION

Balance control

The foam-induced 40% decrease in EQ corroborates our hypothesis on the effect of increased compliance of the supporting surface on CoP displacement. This also establishes the presence of a decrease in EQ as an alternative effect of standing on foam on balance control, i.e., besides a proprioceptive effect. The 40% decrease in EQ means that the efficiency of frontal plane foot tilt, with respect to the generation of CoP displacement, is decreased to 60%. Without changes in the amplitude and velocity of foot tilt, this will result in a 40% reduction in the amplitude and velocity of the CoP. In view of the function of CoP displacement in balance control this is bound to have a negative effect on balance control. However, the increase in A_{Mf} and v_{Mf} show that the body attempts to prevent the impending decrease in the amplitude and velocity of CoP displacement by increasing the amplitude and velocity of foot tilt. The fact that this results in virtually unchanged values for A_X and v_X, suggests that this compensation was successful. This is supported by the fact that F_x, as a measure of balance control, was not significantly changed.

Because Y does not change significantly, it can be assumed there was no substantial change in foot loading, that could affect EQ. The small increase in Y observed would have caused a minor increase in EQ and, consequently, resulted in a slight underestimation of the effect of foam on EQ.

Despite a compensatory increase in the amplitude and velocity of foot tilt, a deterioration of balance control can still occur, namely in the following situations. Firstly, when the subject is unable to generate the increased FTS activity necessary for a complete compensation. This depends on the amount of compensation necessary, the active range of motion of the ankle joint and the capacity to generate fast foot tilt. On the one hand a decrease in EQ can be so large that it is impossible, even for the young and physically fit, to compensate for. On the other hand, it may be impossible for the elderly or disabled to compensate for a (minor) decrease in EQ that can easily be compensated for by young and healthy subjects. A second cause for deteriorated balance control is the fact that faster movements are generally less well coordinated (Fitts 1954; Fitts and Peterson 1964). This means that, although larger and faster foot tilt maintains the amplitude and velocity of CoP displacement, it results in a decrease in the coordination of CoP displacement. This, in turn, has a negative effect on balance control. It can be concluded that the ultimate effect of foam on balance control is, among others, dependent on the amount of EQ decrease and the subject's capacity to compensate for this decrease.

Table 1 indicates that foam has different effects on frontal and sagittal plane balance control. In contrast to the frontal plane, the sagittal plane velocity of CoP displacement was significantly increased. This increase in v_Y can be interpreted as a compensation for a negative effect on balance control. The presence of such an effect is supported by the increase in F_y, reflecting a deterioration of balance control. There are several potential causes for the increases in v_Y and F_y. Firstly, the decrease in A_Y, which could have been the result of some constraining effect on ankle mobility or muscle power, or just coincidental. We cannot substantiate the presence of a constraining effect. Therefore, and because the effect on A_Y was not significant, the possibility of a constraining effect on A_Y was rejected. Secondly, the foam could have a stronger effect on the sagittal plane EQ value. Although there is no method to measure the sagittal plane EQ value, this option cannot be supported by either the RSI model or the SB model, and is therefore rejected. Finally, the effect of foam on frontal and sagittal plane proprioception could be different: this possibility is discussed below.

Proprioception

During stance, two sources of proprioceptive information can be distinguished. In the first place the skin of the foot, and the ligament and capsule of the ankle joint. The working range for balance stabilization of these structures is generally regarded to be below 1 Hz (Diener et al. 1984). A working range below 1 Hz is comparable to the working range of the visual system (Dichgans and Brandt 1973; Diener et al. 1982). This is supported by the finding that loss of afferent information from the aforementioned structures, whether due to ischemia (Diener et al. 1984) or anesthesia (Konradsen et al. 1993) can be compensated for by visual information. This means that, in the presence of visual information, standing on foam will not affect balance control via diminished afferent information from the skin of the foot and the ligament and capsule of the ankle joint. The second source of afferent information is the muscle spindles and Golgi tendon organs in the muscles of the shank. The working range of these structures is above 1 Hz (Goodwin et al. 1976; Poppele and Kennedy 1974). This means that, loss of afferent information from these structures, e.g. due to ischemia up to the level of the thigh, cannot be compensated for by visual information (Mauritz et al. 1980; Diener et al. 1984). In the sagittal plane the supporting foot rests on the floor during unperturbed one-leg stance. In this situation, body tilt changes the length and tension of the muscles in the shank and excitates muscle spindles and Golgi tendon organs. This provides information about the position of the movements and the body. However, when standing on foam, the length and tension of the same muscles are altered by the movements of the body as well as tilting of the foot. This increases the ambiguity of the afferent information and could deteriorate sagittal plane balance control. In the frontal plane, however, foot tilt is omnipresent, even on a solid floor. This suggests that, compared to sagittal plane balance control, afferent information from shank muscles is less important for frontal plane balance control during one-leg stance, on a solid surface. Consequently, standing on foam should have a less deteriorating effect on frontal plane balance control via decreased proprioception.

This is supported by the significant increase in F_y in the absence of an appreciable effect on F_x (Table 1). This suggests that, besides the amount of EQ decrease and the subject's physical capabilities, the ultimate effect of foam on balance control is also determined by the plane in which postural control is studied.

The absence of a negative effect on frontal plane balance control and proprioception found in the present study seem in contradiction with the effects of foam reported by others (Bles and De Wit 1975; Brandt et al. 1981; Di Fabio and Badke 1991; Ledin et al. 1993; Norre 1992 and 1993; Teasdale et al. 1991; Weber and Cass 1993; Yardley et al. 1992). However, there are several differences between their studies and the present one. First, in the cited studies subjects were measured during two-leg stance. The contribution of proprioception to balance control increases with decreased body movements (Diener et al. 1984; Horak et al. 1990), which fits our findings. Compared with two-leg stance, body movements during one-leg stance are more intense. Consequently, proprioceptive information is probably less important during one-leg stance. Therefore, a decline in proprioception is expected to have a stronger impact on balance control during two-leg stance. Second, all cited researchers used foam thicker than 5 cm. This is expected to have a stronger negative effect on balance control. Finally, in the majority of the cited studies elderly subjects or patients with balance disorders were measured (Di Fabio and Badke 1991; Norre 1992 and 1993; Teasdale et al. 1991; Weber and Cass 1993) or subjects were sedated (Ledin et al. 1993). Obviously it is more difficult for this type of subject to compensate for a decrease in proprioception or EQ than for the healthy, young and unconstrained subjects in the present study. This means that, besides the amount of EQ decrease, the subject's physical capabilities and the plane in which postural control is studied, the ultimate effect of foam on balance control is also determined by the type of stance (i.e. one v. two-leg).

CONCLUSIONS

It can be concluded that, besides an effect on proprioception, standing on foam can affect frontal plane balance control by decreasing the efficiency or gain with which foot movements displace the CoP. For one-leg stance this was experimentally confirmed by a decrease in EQ. The ultimate effect of foam on balance control is determined by the amount of efficiency (gain) decrease, the subject's physical capabilities for compensation, the plane in which postural control is studied and the type of stance (one v. two-leg). During one-leg stance, young, healthy subjects can compensate for a 40% decrease in EQ by increasing the amplitude and velocity of foot tilt, without a deterioration of balance control.

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Chapter IV

Variations in foot breadth: Effect on aspects of postural control, during one-leg stance

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List of abbreviations

CoP	center of pressure
CoG	center of gravity
EQ	the efficiency quotient of foot tilt
FFB	functional footbreadth
FTS	foot tilt strategy, a mechanism for balance control
HAT	body segment comprising of head, arms and trunk
RSI	rocker shaped interface with the supporting surface, representing the supporting foot
h_M	height of the lateral malleolus
M	the horizontal, frontal plane position of the lateral malleolus
R	the radius (mm) of the circle of which the curved surface of the RSI is a part
X	the frontal plane position of the CoP relative to the forceplate coordinate system
Y	the sagittal plane position of the CoP relative to the forceplate coordinate system
F_s	compensatory horizontal shear forces
F_{rg}	the vertical ground reaction force acting on the supporting foot
F_n	the horizontal ground reaction force in the frontal (n=x) or the sagittal (n=y) plane
A_X	the median amplitude of the displacement of the CoP in the frontal plane
A_Y	the median amplitude of the displacement of the CoP in the sagittal plane
A_M	the median amplitude of the displacement of the lateral malleolus in the frontal plane
v_M	the median amplitude of the velocity of the lateral malleolus in the frontal plane
v_X	the median amplitude of the velocity of the CoP in the frontal plane
v_Y	the median amplitude of the velocity of the CoP in the sagittal plane

INTRODUCTION

Posture, by definition, refers to the position of the total body or an individual body segment relative to gravity[1]. For postural control we rely on visual, vestibular and somatosensory information as well as on the action of the musculoskeletal system. The study of postural control is relevant for otolaryngology, neurology, orthopedics, sports-medicine and rehabilitation. A much practiced method to study postural control is platform stabilometry. In this method, subjects generally, have to stand on a forceplate, and remain as motionlessly as possible while the vertical, fore-aft and medio-lateral components of the ground reaction force vector are being measured[2, 3]. From these forces several parameters can be calculated that are related to the movements of the subject standing on the forceplate. The most popular parameters used in platform stabilometry are based on the displacement of the center of pressure (CoP), i.e. the point of application of the resultant vertical ground reaction force (F_{rg}), or on the horizontal components of the ground reaction force[2].

In clinical research the amplitude of the displacement of the CoP during one-leg stance is generally regarded as a measure for body sway[4-13]. However, in an earlier paper[14] we proposed and validated a model for the foot that mechanically links the tilting movements of the foot and the displacement of the CoP in the frontal plane, during one-leg stance. This model is developed as part of a project to develop a method to assess the control of fine movements of the lower extremities. This kind of movements predominantly occur in the frontal plane, during one-leg stance. The main reason for the development of this model was the aforementioned misconception that CoP displacement represents the amount of body sway in the frontal plane. With the help of this model it was shown that, during one-leg stance, frontal plane CoP displacement predominantly reflected the tilting movements of the foot as a mechanism for postural control. This mechanism was called the foot tilt strategy (FTS).

The two major mechanisms for frontal plane postural control during one-leg stance are the foot tilt strategy[14] and the hip strategy[15]. These mechanisms imply the presence of rotations about the subtalar and hip joint respectively. Consequently, in the present study we model the human body as consisting of three segments. The superior segment represents the head, arms and trunk (HAT), the middle segment represents the supporting leg between the hip joint and the subtalar joint and the inferior segment represents the supporting foot (Figure 1A). The knee is not included in this model because frontal plane rotations about the extended knee joint are negligible. As is customary in linked segment models, all joints are modelled as hinge joints[16].

The inferior segment, representing the foot caudal to the subtalar joint was modelled as a rocker shaped interface (RSI) with the supporting surface. The linked segment model, including the RSI, hereafter will be referred to as "the RSI model".

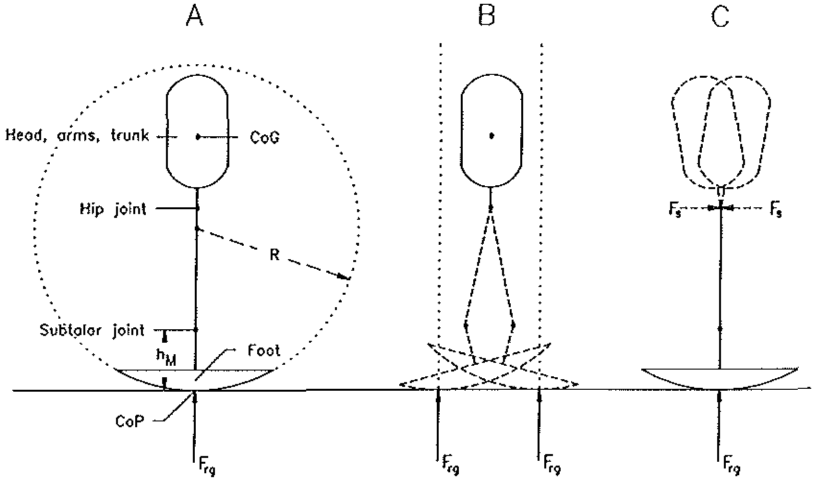

Figure 1. Linked segment model (A) with the supporting foot modelled as a Rocker Shaped Interface (RSI) with the supporting surface. The foot tilt strategy (FTS) causes displacement of the point of application of the vertical ground reaction force (F_{rg}), i.e. the CoP, relative to the body's center of gravity (CoG) (B). The hip strategy, i.e. rotation of the upper body about the hip joint, causes compensatory horizontal shear forces (F_s) (C).

This RSI model has two mechanisms for postural control. First, the FTS in which stabilizing moments of force are generated due to displacement of the ground reaction force relative to the center of gravity (CoG) of the body (Figure 1B). Second, the hip strategy in which stabilizing moments of force are generated due to horizontal shear forces (F_s) as the result of rotation of the upper body about the hip joint[15] (Figure 1C). According to the configuration of the RSI, the relation between the horizontal displacement of the lateral malleolus (ΔM) and the CoP (ΔX) in the frontal plane is given by

$$EQ = \frac{\Delta X \cdot h_M}{\Delta M} \tag{1}$$

the derivation of which was described earlier[14]. In this formula, h_M is the height of the lateral malleolus in mm, at the level of its most lateral point. EQ is called the efficiency quotient of foot tilt and is defined as the displacement of the CoP in mm per radian of foot tilt. The value of EQ is numerically equal to the radius (R) in mm of the circle of which the circumference of the RSI is a part (Figure 1A).

CoP displacement contributes to postural control by continuously moving ahead and behind the body's center of gravity[16]. Therefore, during postural control, demands are being made on the amplitude and velocity of CoP displacement. During one-leg stance the amplitude and the velocity of CoP displacement depend on the amplitude and the velocity of foot tilt and on EQ. This means that a decrease in EQ will cause a decrease in the amplitude and velocity of CoP displacement unless there is a compensatory increase in the amplitude and velocity of foot tilt. Unless this compensation restores the amplitude and velocity of CoP displacement to their previous levels, there will be a decrease in the contribution of the FTS to postural control. This, in turn, results in a deterioration of postural control. An increase in EQ is expected to be easily compensated for by a decrease in the amplitude and velocity of foot tilt. This means that any factor altering, and especially decreasing, EQ could affect postural control. Because it can be inferred from the configuration of the RSI that R reflects the breadth of the foot[14], one of the factors hypothesized to affect EQ was foot breadth. It was predicted by the RSI and experimentally confirmed that there is a strong, linear relation between maximum foot breadth and the value of EQ during erect stance[14]. A narrow or wide foot resulted in a low or high value of EQ, respectively. It was also predicted that intra-subject variations in foot breadth would have a similar effect on EQ. To quantify intra-subject variations in foot breadth we introduce the concept of the functional foot breadth (FFB) defined as the average breadth of the part of the foot predominantly supporting the bodyweight and being used by the FTS for postural control. Because the foot widens from posterior to anterior the FFB will increase when a more anterior part of the foot is loaded. The RSI predicts that an increase in FFB will result in an increase in EQ. In view of the assumed relation between FFB and EQ and the effect that changes in EQ could have on postural control we expect that an increase or decrease in FFB could result in respectively improved or deteriorated postural control. Knowledge about the effects of variations in FFB on postural control is relevant for the study of the effects of foot pathology, foot wear, orthoses and foot size on postural control.

The purpose of this study is to test the hypothesis, based on the RSI configuration that changing the weight of the body toward the narrow rearfoot or the wide forefoot causes respectively a decrease or increase in EQ. Subsequently, we examined the effect of these variations in EQ on postural control.

METHODS AND MATERIALS

The independent (FFB) and dependent (EQ and postural control) variables were operationalized as follows.

FFB

Y represents the position of the CoP in the sagittal plane in millimeters relative to the forceplate coordinate system. The mean value of Y indicates the part of the forceplate loaded by the foot. When the position of the foot is constant, which is the case in platform stabilometry, the mean value of Y also indicates the part of the foot supporting the bodyweight, i.e. being used by the FTS for postural control. Because the foot widens from posterior to anterior and because more anterior loading of the foot will cause an increase in Y, an increase or decrease in Y will be interpreted as respectively an increase or decrease in FFB. To shift foot loading to the narrow heel or the wide forefoot the subjects were instructed to lean backwards or forwards, respectively. This resulted in the following three measurement postures:

1. leaning backwards as much as possible with leg and upper body straight.
2. standing as erect as possible.
3. leaning forwards as much as possible with leg and upper body straight.

EQ

EQ is the efficiency quotient of foot tilt and is defined above. EQ is calculated according to equation 1.

FTS

FTS activity is represented by the average amplitude (A_M) and velocity (v_M) of the frontal plane displacement of the lateral malleolus. These variables reflect the average amplitude and velocity of frontal plane foot tilt. The contribution of FTS activity to postural control is reflected in the average amplitude (A_X) and velocity (v_X) of the frontal plane displacement of the CoP.

Postural control

During the assessment of postural control, subjects have to stand as motionlessly as possible. Postural control is then judged by the amount of residual body movements. The less movements the better postural control is judged to be. Parameters generally used to quantify frontal plane postural control are A_X and F_x. These are both forceplate parameters, i.e. they can be calculated from forceplate data alone. The meaning of A_X was discussed above. F_x is the horizontal, frontal plane component of the ground reaction force and represents the horizontal, frontal plane component of the acceleration of the body's center of mass. Contrary to A_X, F_x takes into account both FTS activity and upper body movements. This explains why CoP based variables have a lower sensitivity and predictive value regarding steadiness compared to force based variables, as was shown by Goldie et al.[2, 3]. As a result of this, we take F_x as the primary variable, normative for postural control.

Although sagittal plane variables cannot be explained in terms of the FTS, because the RSI model is only valid for the frontal plane, they do give information about sagittal plane postural control. To get an impression of the effect of the three measurement postures on sagittal plane postural control, the sagittal plane amplitude (A_Y) and velocity (v_Y) of CoP displacement in respectively mm and mm s^{-1} were measured. We also measured the horizontal component of the ground reaction force in the sagittal plane (F_y) representing the horizontal component of the acceleration of the body's center of mass in this plane.

Subjects

We measured seven men and three women with a mean age of 27 years (range 22 to 37 years), a mean height of 1.78 m (range 1.64 to 1.92), a mean weight of 77.5 kg. (range 55.7 to 102.2) and a mean foot length and breadth of 255 mm (range 230 to 287) and 98 mm (range 90 to 104) respectively. None of the subjects had a history of major orthopedic or neurological disorders or was injured during measurements. All subjects were recruited from the university population.

Instruments

The position of the CoP was measured with a forceplate with piezoelectric force transducers (Kistler type 9281B, Kistler Instrumente AG, Winterthur, Switzerland). The platform was connected to electronic amplifier units (Kistler types 5001, 5006 and 5675).

The movements of the lateral malleolus of the left foot were measured with a displacement transducer (Schlumberger DF\5.0\S, Sangamo transducers, Bognor Regis, England) connected to the left one of two horizontally pivoting flaps between which the ankle is positioned (Figure 2). Because the two flaps of the ankle measuring apparatus were connected by an elastic band, the subject's lateral malleolus was permanently in contact with the flap connected with the displacement transducer. To prevent sagittal plane ankle movements from affecting the frontal plane position of the displacement transducer, the supports on which the flaps were mounted were movable in the sagittal plane. To prevent displacement of the lateral malleolus relative to the left flap, a plastic block with a u-shaped groove, to accommodate the lateral malleolus, was mounted on the inside of the left flap (Figure 2). We measured the left foot because the displacement transducer was mounted on the left side, and because the lateral malleolus is more protruding thus reducing the risk of displacement of the ankle relative to the flap connected with the displacement transducer. In healthy subjects there is no difference in postural control when standing on the left or the right foot[2, 3, 4, 9, 17]

To minimize electronic drifts, the equipment was allowed to temperature stabilize for at least one hour before the start of data acquisition. Forceplate and displacement transducer signals were low-pass filtered with Butterworth type filters (Kemo LTD, Beckenham, England; Krohn-Hite Corp., Avon, Mass., USA) with the cut-off frequency at 25 Hz. Output signals were recorded at 100 samples per second with a data acquisition board (DAS-1602, Keithly Metrabyte, Taunton, MA, USA) and fed into a personal computer for later analysis. Each measurement consisted of 1024 samples resulting in a test time of 10.2 sec.

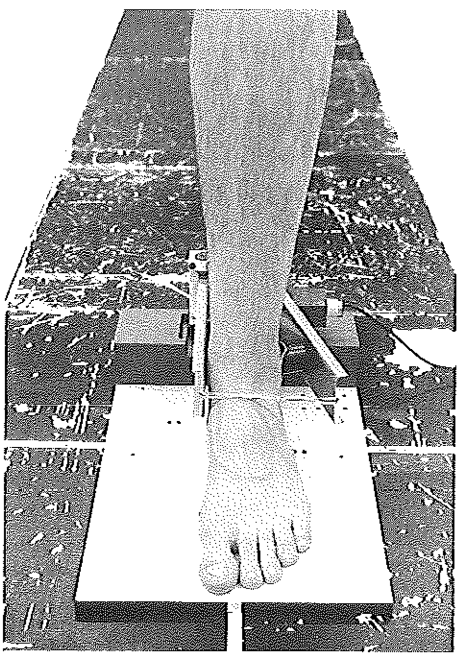

Figure 2. Displacement transducer operated device measuring the horizontal, frontal plane displacement of the lateral malleolus as a measure of foot tilt.

Procedure

After being explained the purpose and the procedure of the experiment, the subjects gave written informed consent and entered the experiment. All subjects were instructed by the same experimenter. The major source of "error" in biomechanical studies on human posture or locomotion is associated with the variability of the subject's performance[18]. In order to minimize intra-subject variability we adopted several measures. During the explanation subjects practiced the three measurement positions. The purpose of this was to verify they understood the instructions. Practicing was also anticipated to improve retest reliability[3]. To prevent a learning effect, the three measurement positions were measured in random order. Two other measures to reduce intra-subject variability were to average the outcome of (three) consecutive measurements[3, 19] and to use a short test time[3]. Our test time (10 s.) was between Goldie's[3] (5 s.) and Geurts's[19] (20 s.). Trials with touchdowns were not accepted because the RSI-model is only valid in one-leg stance. Measurements took place in a well-lighted room. The height of the lateral malleolus was measured with a ruler at the level of its most lateral part.

The position of the supporting foot was: left foot in the center of the forceplate, y-axis through the first interdigital space and the middle of the heel, x-axis through the vertical projection of the medial malleolus on the forceplate, lateral malleolus against a flap of the ankle measuring apparatus. The foot was positioned in the center of the forceplate because the accuracy in determining the position of the CoP is maximal in the center of the forceplate[20]. To prevent their use in postural control, the non-supporting knee was bent about 90° and held against the contralateral knee and both arms were held crossed against the chest. The purpose of this posture was to let the subject approach the model in Figure (1A) as closely as possible in terms of available mechanisms for postural control. Subjects were asked to use a black dot, 24 mm in diameter, attached to the wall 3.6 m in front of them as a visual reference point and stand as motionless as possible. After having assumed the correct measurement position the subjects were allowed 15 sec to settle before the start of data acquisition. Again this is believed to reduce intra-subject variability. Between each measurement the subjects had a one-minute break to prevent fatigue.

Statistical analysis

To compare postures 1, 2 and 3 initially a repeated measures analysis of variance (MANOVA) was conducted with a within-subject design with posture as a within-subject factor and Y, EQ, A_M, v_M, A_X, v_X, F_x, A_Y, v_Y and F_y as dependent variables. When there was a statistically significant effect on a variable, the nature of this effect was further examined with t-tests for paired samples. A p-value $\leq .05$ was considered statistically significant. Statistical analysis was performed with the program SPSS/PC+™ (version 5.02).

RESULTS

According to the repeated measures analysis of variance, measurement position had an effect on all parameters under investigation except v_X (Table 1). As we expected, leaning backward and forward resulted in respectively a decrease and increase in Y. These changes in Y were coupled with changes in EQ. Relative to its value during erect stance, leaning backwards resulted in a 25% decrease in EQ. Similarly, leaning forwards resulted in a 15% increase in EQ. The paired t-tests show that, relative to its value during erect stance, a decrease in EQ is coupled with a consistent and significant increase of all other variables under investigation except of course Y (Table 1). An increase in EQ was coupled with a significant increase in v_Y, a non-significant increase in F_x, F_y, v_M, A_Y, and a non-significant decrease in A_M and A_X (Table 1). The mean value of h_M was 76.6 (SD = 3.8 mm).

The inclination of the body and postural control

Variable$^{\text{MANOVA p-value}}$	Posture$^{\text{T-test p-value}}$		
	Backwards	Erect	Forwards
$Y^{.000}$	$28.36^{.002}$ (SD 13.86)	55.32 (SD 15.98)	$100.52^{.000}$ (SD 17.55)
$EQ^{.000}$	$236.69^{.018}$ (SD 65.46)	321.60 (SD 75.60)	$366.87^{.064}$ (SD 93.65)
$A_X^{.003}$	$5.34^{.015}$ (SD 0.76)	4.38 (SD 0.87)	$4.21^{.565}$ (SD 0.72)
$v_X^{.104}$	23.58 (SD 3.88)	21.01 (SD 4.59)	24.05 (SD 5.91)
$A_Y^{.003}$	$7.31^{.018}$ (SD 3.27)	4.74 (SD 0.99)	$4.80^{.862}$ (SD 1.06)
$v_Y^{.001}$	$29.66^{.005}$ (SD 10.04)	20.44 (SD 5.43)	$24.66^{.047}$ (SD 5.98)
$A_M^{.000}$	$1.86^{.007}$ (SD 0.59)	1.11 (SD 0.41)	$0.95^{.174}$ (SD 0.35)
$v_M^{.001}$	$7.79^{.005}$ (SD 3.26)	4.48 (SD 1.73)	$5.30^{.263}$ (SD 2.42)
$F_x^{.002}$	$2.93^{.008}$ (SD 0.98)	1.99 (SD 0.41)	$2.37^{.098}$ (SD 0.83)
$F_y^{.000}$	$2.35^{.005}$ (SD 0.84)	1.60 (SD 0.36)	$1.63^{.731}$ (SD 0.44)

Table 1. Mean values and standard deviations (SD) of forceplate variables registered during one-leg stance while subjects were: leaning backwards, standing erect or leaning forwards with the supporting knee straight. The variables of interest are: the mean position of the CoP in the saggital plane (Y), the amount of CoP displacement per radian of foot tilt (EQ), the amplitude and velocity of the displacement of the CoP in the frontal (A_X and v_X) and sagittal (A_Y and v_Y) plane, the amplitude and velocity of the horizontal displacement of the lateral malleolus (A_M and v_M) representing the amplitude and velocity of foot tilt and the horizontal component of the reaction force vector in the frontal (F_x) and sagittal (F_y) plane. (n = 10)

DISCUSSION

FFB

The highly significant differences between the mean values of Y in the three measurement postures show that during these postures, different parts of the foot were loaded. This means that by altering the inclination of the body we created three situations in which different parts of the foot were used by the FTS. Because Y increased in the order: leaning backwards, standing erect and leaning forwards, we can expect increasing values for the FFB in this order.

EQ

The consistently increasing values of EQ with increasing values of Y (Table 1) confirm the RSI based hypothesis that EQ changes parallel to the FFB. This means that when the bodyweight is shifted towards the rearfoot, the efficiency of foot tilt, with respect to the generation of CoP displacement decreases. Consequently larger and faster tilting movements of the foot are necessary to maintain the levels CoP displacement and postural control. When, on the other hand, the bodyweight is shifted towards the forefoot, the efficiency of foot tilt, with respect to the generation of CoP displacement increases. This means that smaller and slower foot movements are sufficient to maintain the levels CoP displacement and postural control. Because rearfoot loading had the strongest effect on EQ, this position is also expected to show the strongest effect on postural control.

Postural control

We will subsequently discuss the effect of a decrease and an increase of the amount of CoP displacement per radian of foot tilt (EQ) on the amplitude and velocity of foot tilt (A_M and v_M) and on postural control (F_x and F_y).

Decreased EQ

The 25% decrease in EQ is accompanied by an increase in A_M and v_M (Table 1). This means that the body attempts to prevent the impending decrease in the amplitude and velocity of CoP displacement by increasing the amplitude and velocity of foot tilt. The purpose of this compensation is to maintain the level of postural control. The increase in F_x (Table 1), which reflects a deterioration of postural control, means that the amount of compensation was insufficient. Because both the amplitude and the velocity of CoP displacement were increased, the presence of other factors, than a decrease in EQ, complicating postural control is suspected. An example of such a factor is a decreased coordination of foot tilt because faster movements, e.g. faster tilting movements of the foot, are generally less well coordinated[21, 22]. An alternative explanation is the following. It can be assumed that during erect stance, when the foot and the shank are in their most prevalent positions, the muscles governing foot tilt are at their optimal length for the generation of force. Leaning backwards will alter the foot-shank angle and the length and tension of these muscles. Due to the bell shaped curve of the length-force relationship of muscle[17], this will negatively affect the force generated by these muscles, and make foot tilt more arduous. Crosstalk from the sagittal plane is another possible explanation. However, this is in contradiction with the low correlation between frontal and sagittal plane CoP displacement as well as horizontal forces[14] and the fact that the magnitude of the effects on sagittal and frontal plane postural control is approximately equal (Table 1).

Although it cannot be explained in terms of changed EQ and FTS activity, the increase in A_Y, v_Y and F_y suggests an additional deterioration in sagittal plane postural control. The effect of muscle length and the likelihood of crosstalk from the frontal plane discussed above, could also apply here. An alternative explanation for the effect on sagittal plane postural control is that, when leaning backwards as much as possible one is on the brink of falling. In this situation, postural control is likely to be more difficult than when standing erect, i.e. close to a position of equilibrium. Difficult sagittal plane postural control could also result in a shift in the attention at the expense of frontal plane postural control.

It is clear that a decrease in EQ can compromise postural control. Besides trauma, surgery and congenital malformations, footwear, e.g. shoes with narrow soles, can decrease the FFB and negatively affect postural control. An extreme example of this effect occurs when standing on a skate, when the FFB is reduced to about 2 mm. This creates, as any skater will confirm, a highly unstable situation. Although, with normal foorwear the effect will be less extreme, it should be taken in consideration, especially during the prescription of footwear for the elderly or the disabled.

Increased EQ

As expected, the increase in EQ caused a decrease in the amplitude of foot tilt. Surprisingly however, instead of maintaining the amplitude of CoP displacement this resulted in a decrease of A_X. When this is solely the result of increased EQ, it could mean that postural control was improved and less CoP displacement was necessary. This is however, in clear contradiction with the observed increase in F_x reflecting decreased postural control. An alternative explanation is that the decrease in A_X is the result of a constraining effect on the amplitude of foot tilt. Constrained foot tilt would be more in agreement with the observed deterioration of postural control. The importance of unconstrained foot tilt for adequate postural control is illustrated by the fact that people with a subtalar[23] or a talonavicular[24] arthrodesis, i.e. with limited foot tilt, experience balancing problems when walking on uneven ground. A constraining effect on A_M is also corroborated by the observed increase in the velocity of foot tilt as well as CoP displacement. We also did not expect opposite changes in A_M and v_M because these variables should alter in parallel when their changes are predominantly the result of compensation for changes in EQ. Again this suggests that factors other than changes in EQ played a role in the observed deterioration of postural control. A possible explanation, encompassing the peculiar change in A_M and A_X as well as the changes in other parameters, is the following. Leaning forward causes dorsiflexion of the foot which alters the mechanics of the ankle joint complex and decreases the mobility of the subtalar joint. This will have a constraining effect on A_M and, despite the increase in EQ, on A_X. This resulted in deteriorated postural control. Other possible causes of deteriorated postural control have been discussed above. The explanation for the increase in A_Y, v_Y and F_y due to a decrease in EQ could be applied here too. As predicted on the basis of the smaller effect on EQ, leaning forwards generally had less prominent effects on forceplate variables.

It is expected that, in the absence of any constraining effects on foot tilt, an increase in EQ results in improved or unaltered postural control. Because the value of EQ is linearly related to foot width[14], wearing shoes, which are generally wider than the foot proper, is likely to increase EQ. This is supported by the finding that wearing shoes can increase postural control[25, 26]. This means that choosing the proper footwear can improve postural control and aid in the prevention of falls in the elderly as well as the disabled. When this effect is indeed the result of an increase in EQ, the method described in this paper can aid in the development of adequate footwear by allowing assessment of the effect of different shoes types on EQ.

CONCLUSIONS

It can be concluded that EQ increased parallel to the FFB as predicted by the RSI. Relative to erect stance, both forward and backward leaning resulted in a deterioration of postural control. This negative effect on postural control is probably caused by factors other than changes in EQ, but also related to the foot-shank angle. This means that factors such as foot pathology, foot wear, orthoses and foot size, that can potentially affect EQ or the foot-shank angle, could affect postural control. Because of the coupling of Y with EQ, FFB and the inclination of the body, the absence or presence of changes in Y can be used to exclude or include the presence of the effects on postural control encountered in this investigation, whatever their origin. The EQ related effects of footwear can be assessed, in advance, with the method described in this study.

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Chapter V

The effect of ankle bracing on frontal plane postural control during one-leg stance

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List of abbreviations

CoP	center of pressure
F_n	the horizontal ground reaction force in the frontal (n=x) or the sagittal (n=y) plane
A_X	the median amplitude of the displacement of the CoP in the frontal plane
A_Y	the median amplitude of the displacement of the CoP in the sagittal plane
A_H	the median amplitude of the displacement of the upper body in the frontal plane
v_H	the median amplitude of the velocity of the upper body in the frontal plane
a_H	the median amplitude of the acceleration of the upper body in the frontal plane
v_X	the median amplitude of the velocity of the CoP in the frontal plane
v_Y	the median amplitude of the velocity of the CoP in the sagittal plane
Y	the sagittal plane position of the CoP relative to the forceplate coordinate system
Δv_X	the brace mediated change in the median amplitude of the velocity of the CoP in the frontal plane
Δv_Y	the brace mediated change in the median amplitude of the velocity of the CoP in the sagittal plane

INTRODUCTION

Ankle injuries occur often in sports and, with increasing age, in normal daily life (11). Ten to twenty percent of acute ligament injuries of the ankle result in chronic, lateral ankle joint instability (17). Therefore, in sports, much attention is paid to the prevention of ankle injuries. An important modality in the prophylaxis of ankle injuries is joint protection. One of the most prevalent methods for joint protection is ankle bracing. Ankle braces resist passive inversion of the foot (23) and reduce the frequency of ankle injuries (24, 25). Among the side effects of ankle bracing are a decrease in athletic performance (3) and an effect on postural control. Both positive (8, 14) and negative (2) effects on postural control have been described. Because sports participation warrants a high level of postural control, any factor negatively affecting postural control will decrease sporting performance. As ankle bracing potentially is such a factor, the nature and cause of its effect on postural control needs to be assessed. The acquired knowledge could also aid in the design and development of ankle braces.

In the article of Jerosch et al. (1994), healthy subjects have to keep their balance while standing on a single leg, on soft ground, for one-minute and with open and closed eyes. Postural control was quantified with the number of failures. The positive effect of ankle bracing on postural control was attributed to "additional proprioceptive and exteroceptive sensory perception". An effect of ankle bracing on proprioceptive and exteroceptive sensory perception is certainly possible because ankle bracing was shown to improve the position sense of the ankle joint during angle reproduction tests (6). However, this does not automatically mean an improvement of postural control, since, as will be discussed below, other than proprioceptive effects of ankle bracing also play a role in postural control. As Jerosch et al. state, "their (i.e. the subjects) performance is dependent upon a plethora of variables other than proprioception".

Bennell and Goldie (1994) used a forceplate to examine the effect of ankle bracing on frontal plane postural control, in healthy subjects, during one-leg stance. Their conclusion of a detrimental effect on postural control was based on an increase in the variability of the medial-lateral ground reaction force and the frequency of touchdowns by the non-supporting leg. They did not show a cause for the deterioration of postural control.

Friden et al. (1989) also used a forceplate to examine the effect of ankle bracing on frontal plane postural control during one-leg stance in subjects with acute ankle injury. Without making any assumptions about its cause, they concluded that ankle bracing improved postural control, on the basis of decreased excursions of the center of pressure (CoP), i.e. the point of application of the resultant vertical ground reaction force.

This conclusion is based on the assumption that the movements of the CoP reflect the swaying movements of the body. However, the view that, during one-leg stance, frontal plane CoP displacement represents frontal plane body sway is erroneous (27). Displacement of the CoP contributes to the maintenance of balance and the control of posture (20, 26, 27). During one-leg stance, frontal plane CoP displacement predominantly reflects the frontal plane tilting movements of the foot, as a mechanism for postural control (12). Consequently, the perceived decrease in the amplitude and velocity of CoP displacement actually reflects a decrease in the contribution of foot tilt to postural control. This could be the result of decreased body movements and improved postural control, but also of a mechanical restriction of foot tilt due to ankle bracing. Because, both theoretically (12) and empirically (9, 10), the horizontal ground reaction force is a better measure of frontal plane postural control than CoP based variables, conclusions about the effect on postural control in the study of Friden et al. cannot be established in the absence of these force data.

Foot tilt mediated CoP displacement contributes to frontal plane postural control and implies rotations of the foot about the subtalar joint. Because the purpose of ankle bracing is to restrict the movements of the subtalar joint, we hypothesize a negative effect of ankle bracing on postural control of healthy subjects, during one-leg stance. We also hypothesize that this effect is caused by constraining the amplitude or velocity of CoP displacement. Because the constraining effect of ankle bracing can be expected to increase with increasing foot tilt, we finally hypothesize that the magnitude of this effect is dependent on the amplitude of CoP displacement. The larger the amplitude of CoP displacement, the stronger the constraining effect of ankle bracing. The purpose of this study is to test the three aforementioned hypotheses.

METHODS AND MATERIALS

The various aspects of frontal plane postural control are operationalized as follows. The quality of frontal plane postural control is represented by the magnitude of the medial-lateral ground reaction force (F_x) reflecting the accelerations of the body's center of mass. The lower F_x, the better postural control is judged to be. Because F_x takes into account foot tilt as well as upper body movements we also measured the amplitude (A_H), velocity (v_H) and acceleration (a_H) of the horizontal, frontal plane movements of a reflective marker on the back of the subject, in mm, mm s^{-1} and mm s^{-2}, respectively. The amplitude (A_X) and velocity (v_X) of frontal plane CoP displacement are measured in mm and mm s^{-1}, respectively. To assess the effect of ankle bracing on CoP displacement in view of its hypothesized dependency on A_X, we first reduced the number of samples of A_X from 2048 to 2047 by calculating the mean of every two consecutive samples. Then we arranged all 2047 values of

A_X according to ascending magnitude. The resulting sorting sequence was then also applied to v_X. Each array of A_X was then subdivided in ten sections. The 5th, 15th, 25th - 95th percentile of A_X represent the average value of A_X in the respective sections. The value of v_X for each section was calculated by dividing each array of 2047 values in nine subarrays of 205 values and one subarray (the last) of 202 values. The value of the 50th percentile of each subarray represented the average value of v_X in that subarray. The A_X dependency of the effect of ankle bracing on A_X (ΔA_X) was assessed by examining the difference between the 5th, 15th, 25th - 95th percentile of A_X in the measurements with and without brace.

To get an impression of the effect of ankle bracing on sagittal plane postural control, sagittal plane variables are also mentioned here. F_y is the sagittal plane component of the horizontal ground reaction forces, and represents the horizontal component of the acceleration of the body's center of mass in that plane. A_Y and v_Y are the amplitude and velocity of sagittal plane CoP displacement in mm and mm s^{-1}, respectively. Y is the mean position of the CoP in the sagittal plane and reflects how the bodyweight is distributed over the foot.

Subjects

Data were collected from ten male and eight female subjects with a mean age of 29 years (range 22 to 42), a mean height of 1.78 m (range 1.52-1.96), a mean weight of 75.5 kg (range 56.6-97.2) and a mean foot length and breadth of 260 mm (range 219-290) and 100 mm (range 88-109), respectively. None of the subjects had a history of orthopedic or neurological disorders. All subjects were students or university personnel.

Instruments

We used a forceplate with piezoelectric force transducers (Kistler type 9281B, Kistler Instrumente AG, Winterthur, Switzerland) to measure ground reaction forces. The platform was connected to electronic amplifier units (Kistler types 5001, 5006 and 5675). Forceplate signals were low-pass filtered with a 4th order Butterworth type filter (Kemo LTD, Beckenham, England) with the cutoff frequency at 20 Hz. Output signals were recorded at 100 samples per second using a DAS1602 data acquisition board (Keithly Metrabyte, Taunton, MA, USA) and fed into a personal computer for later analysis. Each measurement consisted of 2048 samples resulting in a test time of 20.5 s. The movements of the reflective marker attached to the back, in the midline at the level of inferior angle of the scapula, were registered with a Sony AVC-D5CE CCD camera (Sony, Japan) connected to a 68030 processor based VME computer system (18). Video signals were recorded with 50 Hz and digitally filtered twice with a second order Butterworth type filter with a final cutoff frequency of 5 Hz (28). For ankle bracing we used an Air-Stirrup ankle brace (Aircast Inc, Summit, NJ, USA).

Procedure

Measurements took place in a well-lighted room. Subjects were asked to use a black dot 24 mm in diameter, attached to the wall 3.6 m in front of them, as a visual reference point and stand as motionlessly as possible. To prevent a learning effect, the measurement order (braced vs. not braced) was determined with restricted randomization. To reduce the effect of intra-subject variability, each subject was measured three times; the mean value of these three repetitions was used in the statistical analysis. The measurement position was: left foot in the center of the forceplate, long forceplate axis through first interdigital space and middle of the heel, short forceplate axis through medial malleolus. The supporting knee was straight but not hyperextended. To prevent their use in postural control, the non-supporting knee was bent approximately 90° and held against the contralateral knee and both arms were crossed and held against the chest. After having assumed the correct measurement position the subjects were allowed 10 s to settle before the start of data acquisition. Between consecutive measurements the subjects had a 30 s break to prevent fatigue. This study conforms to the policy statement regarding the use of human subjects and written informed consent as published by Medicine and Science in Sports and Exercise.

Statistical analysis

The distribution of all variables of interest was examined with normal probability plots and the Shapiro-Wilk test for normality (1). Dependent on the distribution, we assessed the effect of bracing with t-tests for paired samples or Wilcoxon matched pairs signed rank sum tests. A p-value $\leq .05$ was considered statistically significant. The relation between A_X and v_X, and the relation between A_X and the effect of ankle bracing on A_X (ΔA_X) and v_X (Δv_X) were examined with regression analysis with A_X as the independent variable. We successively used a linear and a non-linear approach. In the linear approach the hypothesis is tested that the relation between two variables can be described with a line with an equation of the form
$y = a_0 + a_1 \cdot x$. In the non-linear approach the hypothesis is tested that the relation between two variables can be described with a line of which the (non-linear) equation is determined after visual inspection of the data, to obtain an optimal fit. The coefficient of determination (r^2) and a p-value (only provided with the linear approach) were used as measures of the goodness of fit. The Δv_X curve was created by subtraction of the fitted v_X curves, with and without brace. Statistical analysis was performed with the program SPSS/PC+™ (version 5.02).

RESULTS

One subject was excluded retrospectively because of inadmissible and unaccountable variations in the vertical component of the ground reaction force. Except for A_H, v_H and a_H all variables appeared to come from Normally distributed populations. Ankle bracing resulted in a deterioration of postural control, characterized by an increase in F_x (Table 1).

	Unbraced	Braced	P-value
F_x (N)	2.00 (SD 0.49)	2.18 (SD 0.67)	.040
A_H (mm)	4.42 (SD 1.48)	4.90 (IQR 2.12)	.227
v_H (mm s^{-1})	4.20 (SD 1.16)	4.17 (IQR 1.50)	.246
a_H (mm s^{-2})	2.32 (SD 0.44)	2.42 (IQR 0.57)	.113
A_X (mm)	4.26 (SD 0.73)	4.71 (SD 0.86)	.038
v_X (mm s^{-1})	24.34 (SD 6.04)	24.42 (SD 7.63)	.929
F_y (N)	1.39 (SD 0.42)	1.49 (SD 0.41)	.103
A_Y (mm)	5.48 (SD 1.85)	5.61 (SD 1.08)	.741
v_Y (mm s^{-1})	20.95 (SD 5.09)	22.22 (SD 5.43)	.130
Y (mm)	56.23 (SD 11.66)	53.79 (SD 12.49)	.120

Table 1. The effect of ankle bracing on the horizontal accelerations of the body's center of mass in the frontal (F_x) and sagittal (F_y) plane, the amplitude and velocity of the horizontal movements of the upper body (A_H and v_H) and the center of pressure (CoP) in the frontal (A_X and v_X) and sagittal (A_Y and v_Y) plane and the sagittal plane position of the CoP (Y). N = 17

There was an increase of A_H and a decrease of v_H, both statistically non-significant (n.s.) (Table 1). Ankle bracing caused intensified foot tilt and increased accelerations of the upper body, characterized by an increase in A_X , v_X and a_H, respectively (all n.s.;Table 1). Testing the hypothesis that the relation between A_X and ΔA_X (Figure 1) can be described by a linear equation resulted in values of r^2, a_0 and a_1 of .97, 0.03 and 0.007 (p= .0000), respectively. Testing the hypothesis that this relation can be described by an non-linear equation of the form $y = b_0 + b_1 \cdot e^{(-x/b2)}$ resulted in values for r^2, b_0, b_1 and b_2 of .97, -0.82, 0.89 and -164.83.

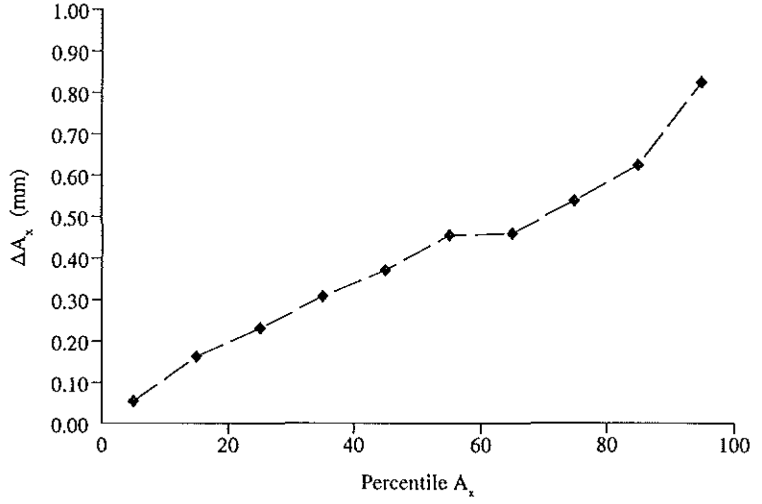

Figure 1. The effect of ankle bracing on the amplitude of CoP displacement (ΔA_X) in the different ranges of A_X.

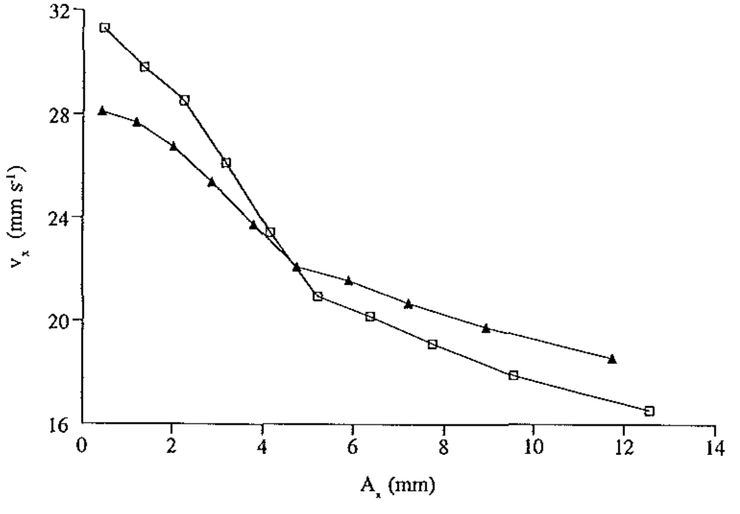

Figure 2. The relation between the amplitude of center of pressure (CoP) displacement (A_X) and the velocity of CoP displacement (v_X) in the situation without (▲) and with (□) ankle brace.

Testing the hypothesis that the relation between A_X and v_X can be described by a linear equation in the situation with and without brace (Figure 2) resulted in values of r^2, a_0 and a_1 of .90, 29.84 and -1.28 (p= .0000) and .93, 27.90 and -0.92 (p= .0000) respectively. In the non-linear approach, testing that the relation between these variables can be described by an equation of the form $y = b_0 + b_1 \cdot e^{(-x/b2)}$ resulted in values for r^2, b_0, b_1 and b_2 of .99, 13.94, 19.33 and 5.71 and .98, 15.95, 13.63 and 6.81 respectively. Testing the hypothesis that the relation between A_X and Δv_X (Figure 3) can be described by a linear equation resulted in values of r^2, a_0 and a_1 of .89, 2.19 and -0.42 (p= .0000), respectively.

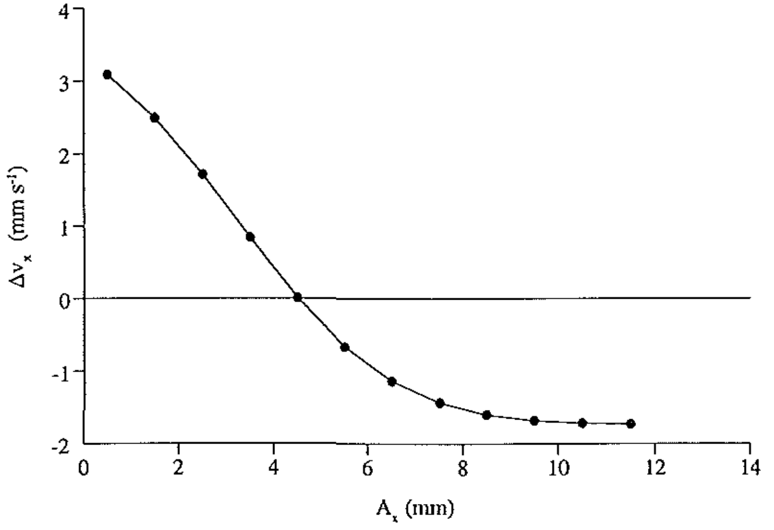

Figure 3. The effect of ankle bracing on the velocity of CoP displacement (Δv_X) with different amplitudes of CoP displacement (A_X).

Testing the hypothesis that this relation can be described by an non-linear equation of the form $y = b_0 + b_1 \cdot e^{(-x/b2)}$ resulted in values for r^2, b_0, b_1 and b_2 of 1.0, -2.37, 6.10 and 4.25. Apart from Y, there was a general tendency of sagittal plane variables to increase due to ankle bracing. None of the effects on sagittal plane variables was statistically significant.

DISCUSSION

The increase in F_x (Table 1) reflects a deterioration of postural control. This finding is comparable with that of Bennell and Goldie (1994) and corroborates the hypothesis that ankle bracing negatively affects postural control. The increases in a_H, A_X and v_X show that intensified movements of both the supporting leg and the upper body contributed to the increase in F_x.

The increases in A_X and v_X (Table 1) are contrary to what we expected and seem to reject our hypothesis about the effect of ankle bracing on CoP displacement. On the one hand an increase in A_X, reflecting increased foot tilt, is in accordance with, and contributes to, the observed increase in F_x. On the other hand it shows that ankle bracing did not have a constraining effect on the average amplitude or velocity of foot tilt. Whether this is due to insensitivity of these variables to the bracing effect or the absence of the hypothesized effect will be discussed later.

The ΔA_X curve (Figure 1) shows that there is no restriction of the amplitude of CoP displacement with increasing values of A_X. On the contrary, the value for a_1 (0.007, p=.0000) shows that there is an increase in A_X in all amplitude ranges. This increase becomes larger with larger amplitudes. This is likely to be the result of deteriorated postural control in the absence of a constraining effect on A_X. Apparently there is enough play between the brace and the foot to allow CoP displacement with increased amplitude.

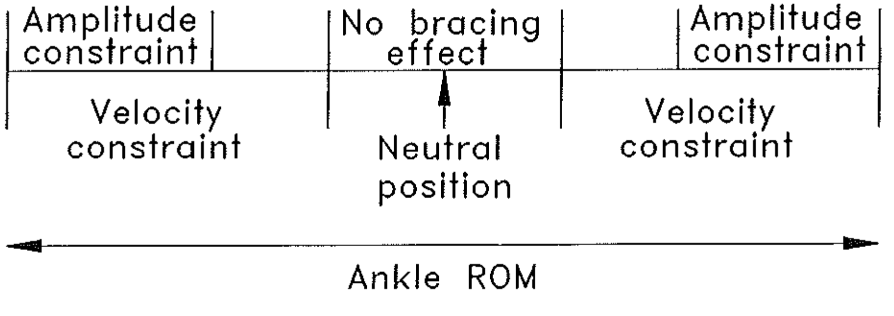

Figure 4. A schematic display of the theoretical levels of constraint of CoP displacement due to ankle bracing.

The course of the v_X (Figure 2) and Δv_X (Figure 3) curves shows that, beyond a certain value of A_X, ankle bracing has a clear, constraining effect on v_X. This effect becomes more prominent with increasing values of A_X. This corroborates both our second and third hypothesis. The increase in v_X in the low amplitude range is likely to be the result of a combination of deteriorated postural control in the absence of a constraining effect on v_X. The above data suggest that, with respect to the constraining effect of ankle bracing on CoP displacement, the range of motion of the subtalar joint can be roughly divided in three ranges (Figure 4). In the range about the neutral position there is no constraining effect on the velocity nor the amplitude of CoP displacement. Beyond this is a range in which there is a constraining effect on the velocity but not on the amplitude of CoP displacement. Finally, there might be a range with the largest amplitudes, where there is a constraining effect on both the velocity and the amplitude of CoP displacement. However, we doubt wether such a range actually exists.

The absence of an effect on the average value of v_X and the course of the Δv_X curve shows that the former variable is of limited value for the characterization of the effect of ankle bracing. Apparently an amplitude dependent approach is necessary for the assessment of the effects of ankle bracing. An amplitude dependent approach is probably also preferable in future analysis of the effects of ankle bracing on A_X in the larger amplitude ranges.

When the results of this study are compared with those of Friden et al. (1989) two important differences can be observed. In the first place both A_X and v_X are considerably larger in our healthy subjects. This is noteworthy because acute ankle injury causes an increase in CoP displacement (5, 8). A possible explanation for this could be the difference in the distance between the subjects and the visual reference point (50 cm vs. 3.6 m) used during the measurements. The closer this point is to the eye, the more effective the visual stabilization of posture is (21). Another possible explanation could be the fact that Friden et al. did not prevent contributions from the arms and non-supporting leg to postural control. The observed dissimilarity could also be the result of a difference in sample frequency (20 Hz vs. 100 Hz). A sample frequency of 20 Hz allows registration of movements up to 10 Hz. This could lead to an underestimation of the higher frequency components of CoP displacement and a decrease in A_X and v_X. A second striking difference between these studies is that opposite effects of ankle bracing on A_X as well as v_X are found. This suggests that ankle bracing has different effects on subjects that are healthy compared to subjects that have acute ankle injury. Possibly the nature of the effect of ankle bracing, in subjects with acute ankle injury is not predominantly mechanical, but more of a proprioceptive nature. The feasibility of such an effect is illustrated by the fact that an increase in peroneal reaction time due to acute ankle injury (16, 19) is partially reversed by ankle taping (16). This effect is attributed to improved proprioceptive input.

Although none of the effects of ankle bracing on sagittal plane variables is statistically significant, the increase in F_y, A_Y and v_Y is in line with a deterioration of postural control. The decrease in Y suggests a change in foot loading towards the narrower rearfoot. Theoretically this could have a negative effect on postural control (13). However, because the effect is minor and not statistically significant we concluded that it does not contribute much to the observed deterioration of postural control.

The effect of ankle bracing described above, indicates a possible difference in the constraining characteristics of ankle braces and ankle ligaments. It can be expected that the constraining effect of an ankle brace or ankle ligaments increases in proportion to the amplitude of ankle motion. When the effect of bracing on v_X is solely the result of this constraint then the resistance against increased foot tilt and Δv_X can be expected to follow roughly the same course. However, the course of the Δv_X curve differs fundamentally from the course of the resistance against passive foot tilt, generated by the ankle ligaments. The effect of ankle bracing on v_X increases fast in the low amplitude range and shows a slower increase at higher amplitudes (Figure 3). The curve of the resistance against passive foot tilt on the other hand, displays a slow initial increase and a fast progression near the extremes of the range of motion (ROM) (4). Whether this discrepancy is the result of a real difference in the constraining characteristics of ankle braces and ankle ligaments, of different effects of ankle bracing on inversion and eversion movements of the foot or the result of active foot tilt control remains, as yet, unclear. The observed course of the Δv_X curve has advantages as well as disadvantages. For healthy subjects a constraining effect early in the ROM seems disadvantageous, because constraint is only needed near the extremes of the ROM. For subjects with ankle injury and increased peroneal reaction time a decrease of v_X early in the ROM is theoretically useful because it gives more time to achieve active ankle bracing by muscle contraction. This suggests that different demands are made upon braces for subjects with and without ankle injury. Both the Δv_X curve and the ligament resistance curve can be characterized by an equation of the form $y = b_0 + b_1 \cdot e^{(-x/b2)}$. The values of b_0, b_1 and b_2 could be used to characterize the bracing characteristics of ankle braces and serve as a guide in the design of ankle braces.

To what extent ankle bracing affects postural control during walking and running is a matter of speculation. The feasibility of a negative effect during walking is illustrated by the fact that people with a subtalar (15) or a talonavicular (7) arthrodesis, i.e. with limited foot tilt, experience balancing problems when walking on uneven ground. The decreased athletic performance due to ankle bracing (3, 22) could very well be an expression of the observed negative effect on postural control.

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Chapter VI

The effect of ankle bracing on frontal plane postural control during one-leg stance on a compliant surface

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List of abbreviations

CoP center of pressure
RoM range of motion
F_n the horizontal ground reaction force in the frontal (n=x) or the sagittal (n=y) plane
A_X the median amplitude of the displacement of the CoP in the frontal plane
A_Y the median amplitude of the displacement of the CoP in the sagittal plane
A_H the median amplitude of the displacement of the upper body in the frontal plane
v_H the median amplitude of the velocity of the upper body in the frontal plane
v_X the median amplitude of the velocity of the CoP in the frontal plane
v_Y the median amplitude of the velocity of the CoP in the sagittal plane
Y the sagittal plane position of the CoP relative to the forceplate coordinate system

INTRODUCTION

The most prevalent sporting injury is a lesion of the lateral ligaments of the ankle. In normal daily life, the number of ankle injuries increases parallel with age (7). Ten to twenty percent of acute ligament injuries of the ankle result in chronic, lateral ankle joint instability (13). Because instability of the ankle joint seriously hampers athletic performance, in sports, there is much attention for joint protection. One of the most popular methods for joint protection is ankle bracing. Ankle braces resist passive inversion of the foot (18) and reduce the frequency of ankle injuries (19, 20). Negative side effects of ankle bracing are a decrease in athletic performance (3, 17) and a deterioration of postural control (2, 10). This negative effect on postural control is the result of a decrease in the velocity of center of pressure (CoP) displacement (v_X) probably due to a constraining effect of ankle bracing on the velocity of foot tilt (10). The CoP is the point of application of the resultant vertical ground reaction force. Because CoP displacement contributes to the maintenance of balance and the control of posture (15, 21, 22), constrained CoP velocity can compromise balance and deteriorate postural control. During one-leg stance, frontal plane CoP displacement is predominantly the result of the tilting movements of the supporting foot (8). Therefore this mechanism for postural control was called the foot tilt strategy (FTS). During undisturbed one-leg stance on a solid surface there is no constraining effect of ankle bracing on the amplitude of CoP displacement (10). Because the constraining effect of ankle bracing can be expected to increase parallel to the amount of foot tilt, a constraining effect on the amplitude of CoP displacement may very well occur in a situation with increased foot tilt. Such a situation can be created by placing a subject on a surface with increased compliance (11). This situation is therefore likely to give more information about the constraining effects of ankle bracing on v_X and the amplitude of CoP displacement (A_X) during increased foot tilt, i.e. nearer to the extremes of ankle range of motion (RoM).

The purpose of this study is to assess the effect of ankle bracing on postural control during a situation with increased foot tilt. We hypothesize a further deterioration of postural control due to a stronger constraining effect on the velocity of CoP displacement. Besides, an additional constraining effect on the amplitude of CoP displacement could occur.

METHODS AND MATERIALS

The various aspects of frontal plane postural control are operationalized as follows. The quality of frontal plane postural control is reflected in the magnitude of the horizontal, frontal plane ground reaction force (F_x). Both theoretically (8) and empirically (5, 6), the horizontal ground reaction force is a better measure for postural control than CoP based variables. F_x reflects the horizontal accelerations of the body's center of mass in the frontal plane. The lower F_x, the better postural control is judged to be. Because F_x takes into account foot tilt as well as upper body movements the latter were also assessed. We measured the amplitude (A_H) and velocity (v_H) of the horizontal, frontal plane movements of a reflective marker on the back of the subject, in mm and mm s^{-1} respectively. The frontal plane amplitude (A_X) and velocity (v_X) of CoP displacement, reflecting frontal plane foot tilt, are measured in mm and mm s^{-1} respectively.

To examine the effect of ankle bracing on CoP displacement in view of its hypothesized dependency on the amplitude of A_X we created arrays of A_X and v_X of equal length. We reduced the number of samples of A_X from 2048 to 2047 by calculating the mean of every two consecutive samples. All 2047 values of A_X were ordered according to ascending magnitude. The resulting sorting sequence was then also applied to v_X. Each array of A_X was then subdivided in ten sections. The 5th, 15th, 25th - 95th percentile of A_X represent the average value of A_X in the respective sections. The value of v_X for each section was calculated by dividing each array of 2047 values in nine subarrays of 205 values and one subarray (the last) of 202 values. The value of the 50th percentile of each subarray represented the average value of v_X in that subarray and the average v_X for the corresponding amplitude range.

To get an impression of the effect of ankle bracing on sagittal plane postural control, sagittal plane variables are also mentioned here. F_Y is the sagittal plane component of the horizontal ground reaction forces, and represents the horizontal component of the acceleration of the body's CoM in that plane. A_Y and v_Y are the amplitude and velocity of the sagittal plane CoP displacement in mm and mm s^{-1} respectively. Y is the mean position of the CoP in the sagittal plane and reflects the average, sagittal plane distribution of the bodyweight over the foot.

Subjects

Data were collected from nine male and seven female subjects with a mean age of 27 years (range 22 to 40), a mean weight of 75 kg (range 52 to 91), a mean length of 1.78 m (range 1.52 to 1.96) and a mean foot width of 98 mm (range 88 to 110). None of the subjects had a history of major orthopedic or neurological disorders. All subjects were students or university personnel.

Instruments

Ground reaction forces were measured with a forceplate with piezoelectric force transducers (Kistler type 9281B, Kistler Instrumente AG, Winterthur, Switzerland). The platform is connected to electronic amplifier units (Kistler types 5001, 5006 and 5675). Forceplate signals were low-pass filtered with a 4th order Butterworth type filter (Kemo LTD, Beckenham, England) with the cutoff frequency at 20 Hz. Output signals were recorded at 100 samples per second using a DAS1602 data acquisition board (Keithly Metrabyte, Taunton, MA, USA) and fed into a personal computer for later analysis. Each measurement consisted of 2048 samples resulting in a test time of 20.5 s. The position of the reflective marker attached to the back, in the midline at the level of the inferior angle of the scapula, was registered with a Sony AVC-D5CE CCD camera (Sony, Tokyo, Japan) connected to a 68030 processor based VME computer system (14). Video signals were recorded with 50 Hz and digitally filtered twice with a second order Butterworth type filter with a final cutoff frequency of 5 Hz (23). For ankle bracing we used an Air-Stirrup ankle brace (Aircast Inc, Summit, NJ, USA). As a compliant surface we used a 5 cm thick layer of foam with a specific weight of 400 N/m^3 (40 $g/dm^{3)}$. Maximum foot breadth was measured with a device in which infrared light emitting diodes and infrared sensors move parallel to each other along the length of the foot. The maximum breadth and breadth of the foot is measured with an accuracy of 1.0 mm.

Procedure

After they had given written informed consent the subjects entered the experiment. Measurements took place in a well-lighted room. Subjects were asked to use a black dot 24 mm in diameter, attached to the wall 3.6 m in front of them, as a visual reference point and stand as motionlessly as possible. The measurement order (braced vs. not braced, foam vs. solid surface) was determined with restricted randomization. To reduce intra-subject variability, each subject was measured three times; the mean value of these three repetitions was used in the statistical analysis. The measurement position was: left foot in the center of the forceplate, long forceplate axis through first interdigital space and middle of the heel, short forceplate axis through medial malleolus. The supporting knee was straight but not hyperextended. To prevent their use in postural control, the non-supporting knee was bent approximately 90° and held against the contralateral knee and both arms were crossed and held against the chest. After having assumed the correct measurement position the subjects were allowed 10 s. to settle before the start of data acquisition. Between consecutive measurements the subjects had a 30 s. break to prevent fatigue.

Statistical analysis

The distribution of all variables of interest was examined with normal probability plots and Shapiro-Wilk tests for normality (1). Dependent on the distribution, we assessed the effect of bracing with t-tests for paired samples or Wilcoxon matched pairs signed rank sum tests. A p-value $\leq .05$ was considered statistically significant. Statistical analysis was performed with the program SPSS/PC+™ (version 5.02).

RESULTS

On a solid surface, ankle bracing resulted in an increase in F_x, F_y, A_X, A_Y, A_H, v_X, v_Y and v_H. Only Y was decreased (Table 1). Apart from the changes in F_y, v_Y and A_X these effects were not statistically significant. There was a trend towards increased F_x and A_Y. Bracing resulted in an increase of A_X in all amplitude ranges (Figure 1). From values of A_X of about 4 mm, ankle bracing resulted in a decrease in v_X (Figure 2).

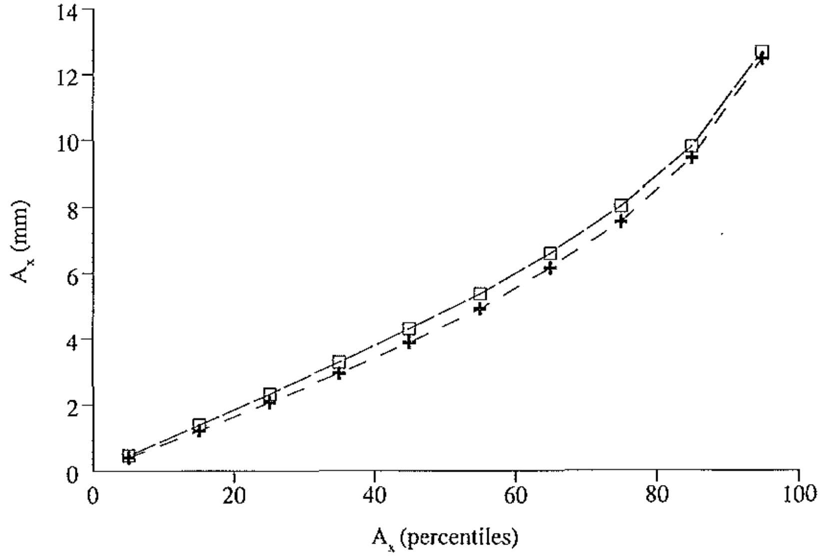

Figure 1. The amplitude of CoP displacement (A_X) without (+) and with (□) an ankle brace on the supporting foot, while standing on a solid surface.

	Solid surface			Foam surface		
	Unbraced	Braced	p-value	Unbraced	Braced	p-value
F_x	2.37(SD .89)	2.60 (SD 1.02)	.058	2.67 (SD .97)	2.71 (SD .84)	.704
A_H	4.90 (SD 2.28)	5.19 (SD 1.48)	.425	4.91 (SD 2.0)	5.82 (SD 2.87)	.081
v_H	5.11 (SD 2.65)	5.56 (SD 2.37)	.291	5.43 (SD 2.0)	6.28 (SD 2.60)	.078
A_X	4.29 (SD .99)	4.85 (SD 1.22)	.001	4.78 (SD .90)	4.92 (SD 1.02)	.484
v_X	25.72 (SD 6.13)	25.98 (SD 6.85)	.971	28.41 (SD 7.68)	25.73 (SD 5.96)	.060
F_y	1.62 (SD .60)	1.77 (SD .67)	.026	1.70 (SD .51)	1.76 (SD .58)	.339
A_Y	4.82 (SD 1.39)	5.34 (SD 1.19)	.090	5.71 (SD 1.44)	5.44 (SD 1.16)	.513
v_Y	23.02 (SD 6.16)	25.42 (SD 7.05)	.003	25.15 (SD 7.23)	26.26 (SD 7.41)	.271
Y	61.88 (SD 13.65)	60.31 (SD 15.41)	.250	62.59 (SD 13.03)	61.03 (SD 13.10)	.227

Table 1. The effects of ankle bracing and standing on 5 cm foam on the horizontal, frontal and sagittal plane ground reaction force (F_x and F_y), the amplitude (A) and velocity (v) of: the frontal plane displacement of the center of pressure (CoP) (A_X and v_X), the sagittal plane displacement of the center of pressure (A_Y and v_Y), the horizontal, frontal plane displacement of the upper body (A_H and v_H), and the average sagittal plane CoP position (Y), N=16.

On a compliant surface however, a completely different picture emanates. In this situation ankle bracing resulted in an increase in F_x, F_y, A_X, v_Y, A_H and v_H. There was a decrease in A_Y, v_X and Y (Table 1). Contrary to the solid surface situation, none of these effects was statistically significant, although there was a trend towards decreased v_X and increased A_H and v_H. Like in the solid surface situation, ankle bracing on a compliant surface resulted in an increase of A_X in all amplitude ranges (Figure 3). Unlike in the solid surface situation, ankle bracing on a compliant surface resulted in a decrease in v_X in all amplitude ranges of A_X (Figure 4).

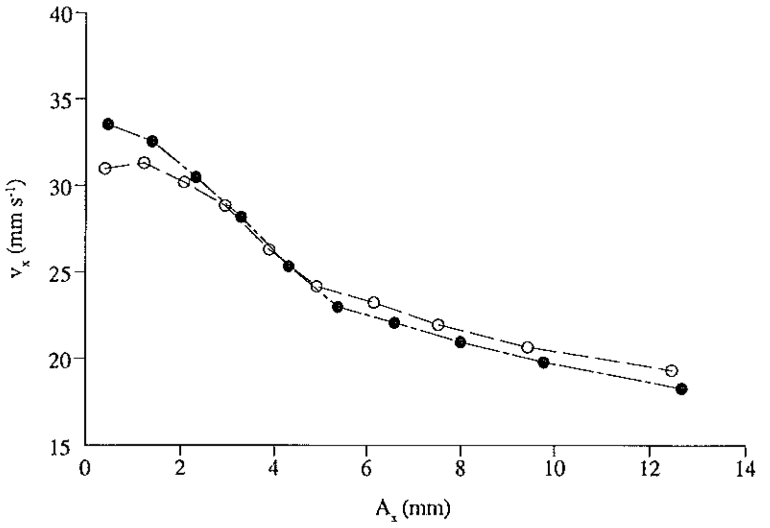

Figure 2. The velocity of CoP displacement (v_X) without (○) and with (●) an ankle brace on the supporting foot, while standing on a solid surface.

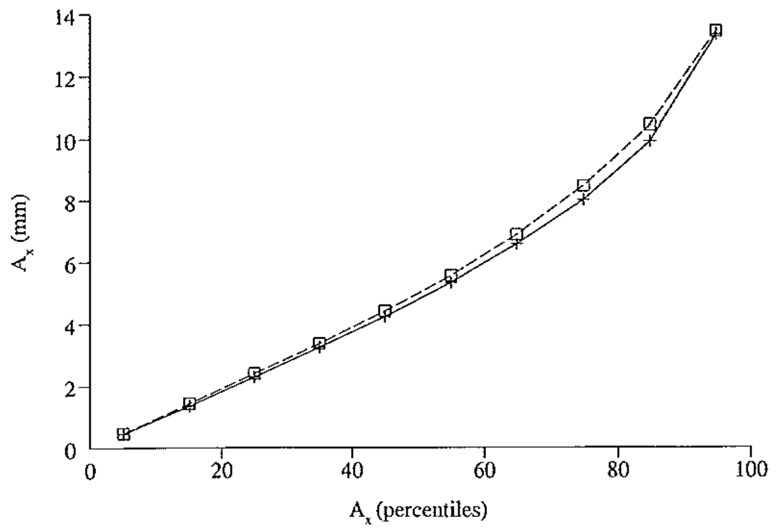

Figure 3. The amplitude of CoP displacement (A_X) without (+) and with (□) an ankle brace on the supporting foot, while standing on 5 cm foam.

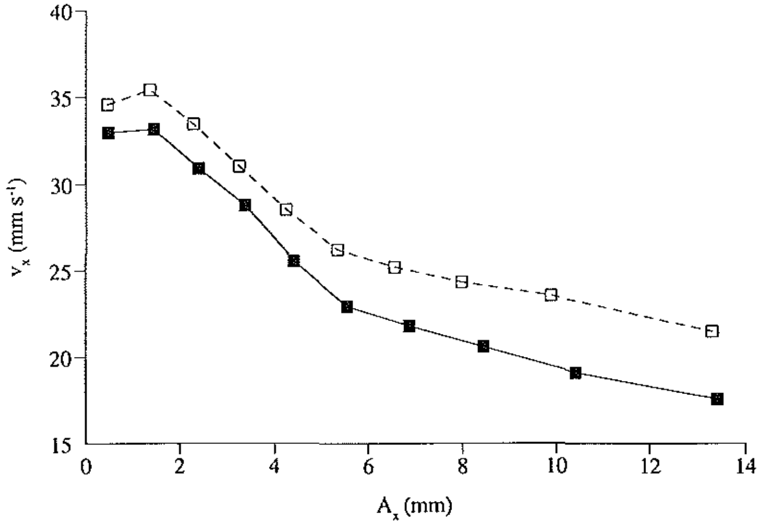

Figure 4. The velocity of CoP displacement (v_X) without (□) and with (■) an ankle brace on the supporting foot, while standing on 5 cm foam.

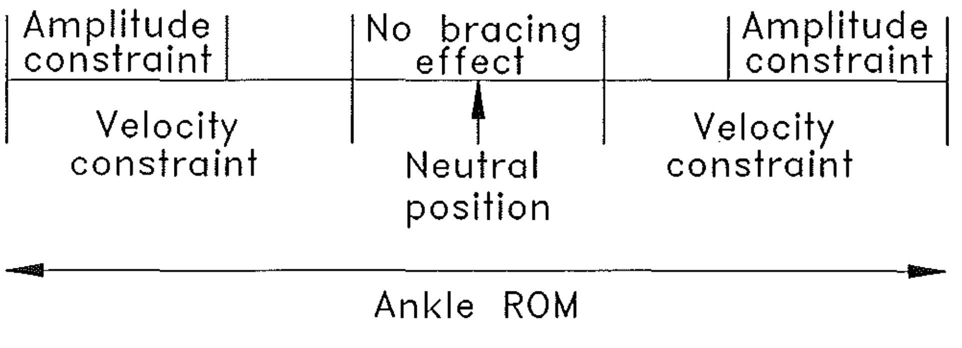

Figure 5. A schematic reproduction of the different levels of constraint of CoP displacement due to ankle bracing.

DISCUSSION

The observed effects of ankle bracing while standing on a solid surface confirm our earlier findings (10): an increase in F_x, i.e. a deterioration of frontal plane postural control, and A_X, and a decrease in v_X concentrated in the higher values of A_X (Figure 1).

The effects of ankle bracing on a compliant surface, however, show a different picture. In the first place, on foam the increase in F_x due to ankle bracing is smaller than in the solid surface situation (Table 1). Because we expected the foam to cause a stronger deterioration of postural control, i.e. a stronger increase in F_x, this finding is contrary to what we expected, and does not corroborate our first hypothesis. A possible explanation for this is that the combination of ankle bracing and foam support did not result in increased foot tilt. However, this argument can easily be refuted. Standing on a 5 cm thick layer of foam with a specific weight of 40 g/dm^3, decreases the amount of CoP displacement per unit of foot tilt by approximately 35% (11). Because of this decreased efficiency of foot tilt, combined with an increase in A_X (Table 1), foot tilt must have been increased. Consequently the constraining effect of ankle bracing must have been increased. This seems to be supported by the stronger effect of ankle bracing on v_X (Table 1) and the course of the v_X curves (Figure 2 and 4). Another possible reason for the smaller than expected increase in F_x is that the combination of ankle bracing and foam causes a shift from the FTS towards the hip strategy as the main mechanism for postural control. In the hip strategy postural control is regulated by rotation of the upper body about the hip joint (12). Such a shift may occur when the constraining effect on foot tilt is too strong to maintain an adequate level of postural control by FTS activity alone. A shift towards the hip strategy would decrease the negative effect of constrained CoP displacement on postural control. A shift towards increased hip strategy is supported by the increase in A_H and v_H (Table 1). A decrease in the importance of the FTS would also explain the relatively small increase in A_X on a compliant surface (Table 1). On a solid surface, ankle bracing results in a strong increase in A_X, most likely to compensate for the decrease in v_X. On a compliant surface however, despite the presence of a stronger constraint of v_X, there is only a minor increase in A_X. An alternative explanation for the minor increase in A_X is a constraining effect of ankle bracing on A_X. However, this argument is refuted by the finding that apart from the 85th percentile the effect on A_X is stronger in the solid surface measurements (Figure 1 and 3). This means that the minor increase in A_X is also present in the lower amplitude ranges, where no constraining effect on A_X is expected.

In the second place, there is a difference in the course of the v_X curves on a solid (Figure 2) and a compliant (Figure 4) surface. Figure (2) shows that on a solid surface a constraining effect on v_X exists from $A_X \approx 4$ mm. On foam there is a decrease in v_X in all amplitudes, although the effect is still strongest in the larger ones. When the constraining effect on v_X was solely amplitude dependent, in the compliant surface measurements a constraining effect on v_X would be expected from $A_X \approx 2.5$ mm ($4 \cdot 0.65$). In the compliant surface measurements a constraining effect on v_X is visible from $A_X \approx 1$ mm. The fact that a constraining effect on v_X is present from lower amplitudes suggests that another factor besides ankle bracing plays a role in the effect on v_X. A possible explanation is offered by the course of the v_X curves in figure (2) and (4). The course of these curves is nearly identical, except from a vertical shift of the curves with and without brace on foam, relative to each other. Again, this phenomenon can be explained by a shift from the FTS towards the hip strategy, causing a general decrease in the velocity of foot tilt. This amplitude independent effect on v_X suggests that the body cannot simultaneously improve the performance of both the FTS and the hip strategy, but improves one at the expense of the other. It was already known that combining a balance related task with a non-balance related task goes at the expense of the former (4). Our data suggests that also during the execution of two balance related tasks the performance for the individual tasks is not independent. This is called structural interference and seems to be the result of a central bottleneck in the sharing of processing resources. Although the increased upper body movements suggest an increase in the activity contribution of the hip strategy to postural control this has not been proven. It would facilitate the interpretation of forceplate data if the activity of the hip strategy could be assessed separately from the activity of the FTS.

In an earlier paper (10) we divided the range of motion of the subtalar joint into three ranges (Figure 5). Near the neutral position there is no constraining effect on the velocity nor the amplitude of CoP displacement. Beyond this is a range in which there is a constraining effect on the velocity but not on the amplitude of CoP displacement. Finally, possibly a range exists with (large) amplitudes and a constraining effect on both the velocity and the amplitude of CoP displacement. According to our data, even when standing on 5 cm foam, no constraining effect of ankle bracing on A_X is discernable. Consequently, on a solid surface, a combination of constrained velocity and constrained amplitude will, on the average, occur beyond $A_X \approx 20$ mm (13 mm / 0.65).

Contrary to solid surface measurements, none of the effects of ankle bracing on sagittal plane variables is statistically significant on foam. The decrease in A_Y reflects decreased activity of the ankle strategy which is in line with a shift towards sagittal plane hip strategy. The increase in F_Y, A_Y and v_Y are also smaller than in solid surface measurements. The decrease in Y suggests a change in foot loading towards the narrower rearfoot. Theoretically this could have a negative effect on postural control (9).

However, because the effect is minor and not statistically significant we concluded that it does not contribute much to the observed effects on postural control.

It can be concluded that the results of the measurements on a solid surface confirm the negative effect of ankle bracing on balance control found earlier. The results of the measurements on a compliant surface suggest a shift from the FTS towards the hip strategy. However, in the absence of knowledge about the activity of the hip strategy this could not be established unambiguously.

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Chapter VII

Effect of body weight and foot size on the generation of frontal plane CoP displacement on a supporting surface with increased compliance

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List of abbreviations

AoS area of support
CoP center of pressure
Eq_{mn} the efficiency quotient of foot tilt, calculated on the basis of A_X and A_M (n=A) or with the slope of the regression line through X as a function of M (n=R), on a solid surface (m=1), on 5 cm foam (m=2) or on 10 cm foam (m=3)
FFB functional footbreadth
b_f the breadth of the supporting foot
$b_{f\text{-}max}$ maximum foot breadth
$l_{f\text{-}max}$ maximum foot length
φ angle of frontal plane foot tilt
h_M the maximum height of the lateral malleolus
K the spring constant or stiffness of the springs, reflecting foam compliance
M the horizontal, frontal plane position of the lateral malleolus
X the frontal plane position of the CoP relative to the forceplate coordinate system
Y the sagittal plane position of the CoP relative to the forceplate coordinate system
A_x the median amplitude of the displacement of the CoP in the frontal plane
A_M the median amplitude of the displacement of the lateral malleolus in the frontal plane
F_{rg} the vertical ground reaction force acting on the supporting foot
p_f the average pressure under the foot

INTRODUCTION

Platform stabilometry is a widely used method for the assessment of postural control. In this method the subject under investigation attempts to maintain its balance while standing on a forceplate that measures the vertical, fore-aft and medio-lateral components of the ground reaction forces. From these forces several variables reflecting aspects of the movements of the subject standing on the forceplate can be calculated. Some of the most often used variables in platform stabilometry are based on the displacement of the center of pressure (CoP), i.e., the point of application of the resultant vertical ground reaction force (F_{rg}). During one-leg stance the amplitude of frontal plane CoP displacement is often used as a measure of body sway[1-10]. However, it was shown that frontal plane CoP displacement predominantly reflects frontal plane foot tilt as a mechanism for postural control[11]. Both the amplitude and the velocity of CoP displacement are important factors in postural control. The importance of the amplitude of CoP displacement for postural control is illustrated by the finding that people with a subtalar[12] or a talonavicular[13] arthrodesis, i.e. with limited foot tilt, experience balancing problems when walking on uneven ground. An example of the importance of CoP velocity is the finding that prophylactic ankle bracing deteriorates postural control by decreasing the velocity of CoP displacement[14].

In view of the above, the efficiency with which the CoP is displaced is an important factor in postural control. During one-leg stance the efficiency quotient of frontal plane foot tilt (EQ) can be calculated with the equation

$$EQ=\frac{\Delta X \cdot h_M}{\Delta M} \tag{1}$$

the derivation of which was described earlier[11]. In this formula, EQ is the amount of CoP displacement in mm, per radian of foot tilt and h_M is the height of the lateral malleolus in mm, at the level of its most lateral point. ΔX and ΔM represent the horizontal, frontal plane displacement in mm, of the CoP and the lateral malleolus, respectively. To calculate the average value of EQ, the average displacement of the CoP (A_X) and the lateral malleolus (A_M) are entered in equation (1) instead of ΔX and ΔM. EQ can be calculated in two different ways. In one way the crude values for A_X and A_M are entered in equation (1). In the other way the slope of the regression line through the values of ΔX as a function of ΔM is entered in equation (1) as the quotient $\Delta X/\Delta M$. The resulting variants of EQ are called EQ_A and EQ_R respectively. Theoretically the effect of these different calculations on EQ is as follows; ΔX

as a function of ΔM follows an ellipsoid course[11]. This course is such that the main non-linearity, i.e., CoP displacement without ankle displacement, occurs at the turning points of the ellipsoid. CoP displacement at these turning points was attributed to forefoot movements independent of foot tilt, and is probably the result of movements of the first ray. The presence of such a mechanism for CoP displacement is corroborated by the activity of the extensor and flexor halluces longus muscles in the middle of the single leg stance phase in walking[15]. In formula (1) an isolated increase in ΔX will increase the value of EQ. By taking the slope of the regression line between ΔX and ΔM for the quotient $\Delta X/\Delta M$ the effect of the turning points on EQ is largely left aside. As a result of this, the values of EQ_R should be consistently lower than the values for EQ_A.

The value of EQ is decreased considerably when one stands on a sheet of foam, i.e., when the compliance of the supporting surface is increased[16]. Foam surfaces are often used in balance research to stress the system governing postural control[17-25]. The effect of a foam surface on EQ can be modeled according to the following equation

$$\frac{\Delta X}{\Delta \varphi} = EQ = \frac{2 \cdot K \cdot b_f^2}{F_{rg}} \tag{2}$$

the derivation of which was described earlier[16]. In this equation $\Delta\varphi$ is the angle of foot tilt, $2b_f$ is the width of the foot and K is the spring constant or stiffness of the springs that represent the foam. According to equation (2) there is a quadratic relation between foot width and EQ. This seems in contrast with earlier findings[11]. Equation (2) also suggests that a thicker layer of foam, with the same specific weight, which results in a decrease in K, will cause a further decrease in EQ. Due to the more than linear increase of the resistance of the foam during compression[26], the value of K increases with increasing compression of the foam. When this is true, the ultimate effect of foam on EQ also depends on the pressure under the foot, i.e., on bodyweight and foot size. To be able to estimate the effects of foot size, body weight and foam thickness on postural control in the individual subject, assessment of the effects of these factors on EQ is necessary. Knowledge about the aforementioned effects on EQ is also necessary to include foot size, body weight and foam thickness in future modeling of postural control.

The aim of this study is to assess the effects of various factors on EQ. First we calculated EQ in two different ways and discussed the resulting differences. Subsequently we examined the relation between foot breadth and EQ. Then we assessed the effect of foam of two different thicknesses on EQ. Finally we assessed the effect of the pressure under the foot on the foam mediated changes in EQ. To obtain a wide range of foot-sizes, a number of children were included in the measurements.

METHODS

In view of the congruence of feet in general we took the maximum foot breadth ($b_{f\text{-}max}$) as a measure for the functional foot breadth (FFB). The FFB is defined as the average breadth of the part of the foot predominantly supporting the bodyweight and being used for balance control. The pressure under the foot was assessed as follows. First we estimated the area of support (AOS), i.e., the area enveloped by the most peripheral points of application of foot to ground force, by tracing the perimeter of the foot. The resulting figure of the foot was clipped out and weighed on an electronic laboratory balance (Sartorius 1265 MP, Sartorius, Nieuwegein, the Netherlands). Because we knew the weight and the dimensions of the original sheet of paper, the area of the clipped out figure could be determined. An alternative method to measure the AOS is based on $b_{f\text{-}max}$ and the maximum length ($l_{f\text{-}max}$) of the foot. These variables are measured with a device (MMS-18, Hoogstraat Techniek B.V., Kampen, the Netherlands) in which infrared light emitting diodes and infrared sensors move parallel to each other along the length and breadth of the foot. Devices of this type are commercially available and are often being used in shoe shops. Length and breadth of the foot are measured with an accuracy of 1.0 mm. We examined the possibility of calculating the AOS on the basis of the maximum length and width of the foot by assessing how well the relation between $l_{f\text{-}max}$ and $b_{f\text{-}max}$ could be described with an equation of the form

$$AOS = a_0 \cdot l_{f\text{-}max} \cdot b_{f\text{-}max} \qquad (3)$$

with a_0 as a weighing coefficient.

When, during one-leg stance, foot loading is altered towards the wide forefoot or the narrow rearfoot this results in respectively an increase and a decrease in the FFB and consequently in EQ[27]. Changing foot loading towards anterior or posterior also results in respectively an increase and decrease in the sagittal plane position of the CoP (Y). The mean value of Y was calculated to detect changes in foot loading that could have affected EQ. Y is expressed in mm, relative to the forceplate coordinate system.

Subjects

We measured 10 male and 14 female subject with a mean age of 24.9 years (range 4 to 46), a mean height of 1.61 m (range 1.01 to 1.96), and a mean weight of 62.2 kg (range 15.0 to 99.3). Foot size was characterized by a mean foot length of 236 mm (range 164 to 291), a mean foot width of 92 mm (range 63 to 114), and a mean foot surface of 169 cm^2 (range 84 to 233). None of the subjects had a history of major orthopedic or neurological disorders. All adult subjects were students or university personnel.

Instruments

For the determination of the position of the CoP we used a forceplate with piezoelectric force transducers (Kistler type 9281B, Kistler Instrumente AG, Winterthur, Switzerland). The platform is connected to electronic amplifier units (Kistler types 5001, 5006 and 5675).

The position of the lateral malleolus of the left foot was marked with a reflective marker (Figure 1).

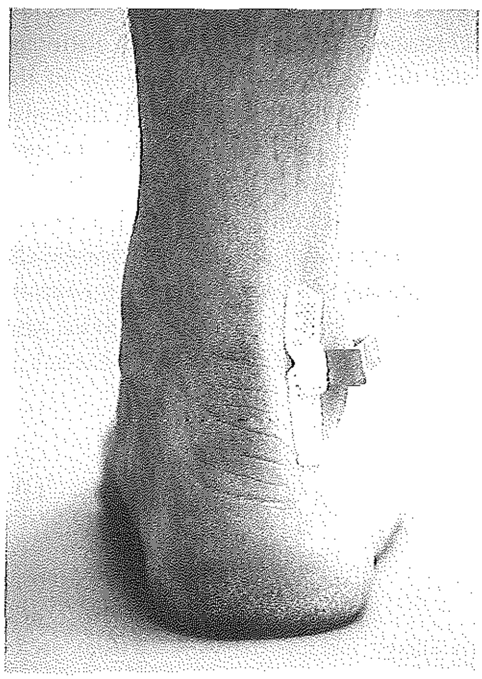

Figure 1. The supporting foot with a reflective marker over the most lateral point of the lateral maleolus.

The movements of this marker were registered with a Sony AVC-D5CE CCD camera (Sony, Tokyo, Japan) connected to a 68030 processor based VME computer system[28]. Video signals were recorded with 50 Hz and digitally filtered twice with a second order Butterworth type filter[29] with a final cut-off frequency of 20 Hz. Forceplate signals were low-pass filtered with a Butterworth type filter (Kemo LTD, Beckenham, England) with the cut-off frequency at 20 Hz. Output signals were recorded at 100 samples per second with a data acquisition board (DAS-1602, Keithly Metrabyte, Taunton, MA, USA) and fed into a personal computer for later analysis. Each measurement lasted 20 s. The solid surface consisted of a 40 mm thick plate of fibreboard, mounted on top of the forceplate. The compliant surfaces consisted of 5 and 10 cm thick sheets of foam with a specific weight of 400 N m^{-3} (40 g dm^{-3}) mounted on top of the fibreboard. In the present study, EQ on the solid surface, on 5 cm foam and 10 cm foam is referred to as EQ_1, EQ_2 and EQ_3 respectively.

Procedure

After they had been briefed about the purpose and the procedure of the experiment, the subjects gave written informed consent and entered the experiment. Measurements took place in a well-lighted room. To prevent a learning effect, measurement order was determined with restricted randomization. Subjects were asked to use a black dot 24 mm in diameter, attached to the wall 3.6 m in front of them, as a visual reference point and stand as motionlessly as possible. To reduce the effect of variability, each subject was measured three times if possible; the mean value of these three repetitions was calculated and used in subsequent statistical analysis. The measurement position was: left foot in the center of the forceplate, forceplate y-axis through first interdigital space and middle of the heel, forceplate x-axis 8.5 cm in front of the rear of the foot. We measured the left foot because of uniformity with earlier studies. In healthy subjects there is no difference in postural control when standing on the left or the right foot[1, 30-33]. The supporting knee was straight but not hyperextended. To increase the chance that every subject completed the measurements, the position of the non-supporting leg and both arms was not restricted. After having assumed the correct measurement position the subjects were allowed unlimited time to settle. Data acquisition started after a verbal sign of the subject. Between consecutive measurements, subjects had a 30-sec break to prevent fatigue.

Statistical analysis

The distribution of the data was assessed with Shapiro-Wilk tests and normal probability plots[34]. Dependent on the distribution, the effect of the method of calculation on EQ and the effect of the compliance of the supporting surface on EQ was examined with paired t-tests or Wilcoxon matched pairs signed rank sum tests. In the case of multiple comparisons a Bonferroni procedure was applied. The relation between $b_{f\text{-}max}$ and EQ was examined with regression analysis with $b_{f\text{-}max}$ as the independent variable. The quotient of bodyweight and AOS was taken as the average pressure under the foot (p_f). The relation between p_f and the foam mediated effect on EQ_R was assessed with regression analysis with p_f as the independent variable. In every case of regression analysis both a linear and a non-linear approach was taken. Unless stated otherwise, in the linear approach the hypothesis is tested that the relation between the dependent and the independent variable can be described with an equation of the form $y = a_0 + a_1 \cdot x$. In the non-linear approach the hypothesis is tested that the relation between the dependent and the independent variable can be described with a non linear equation. To obtain an optimal description the form of the equation is determined on the basis of a graphical display of the data. The coefficient of determination (r^2) and a p-value (only provided with the linear approach) were used as measures of the goodness of fit. In all cases a p-value $\leq .05$ was considered statistically significant. Statistical analysis was performed with the program SPSS/PC+™ (version 5.02).

RESULTS

Apart from EQ_{1A}, EQ_{1R}, EQ_{2A} and EQ_{2R} all variables come from normally distributed populations. There is a significant difference between EQ_{1A} (366.0, IQR 260.9) and EQ_{1R} (258.3, IQR 122.4), between EQ_{2A} (225.9, IQR 139.3) and EQ_{2R} (171.9, IQR 111.2) and between EQ_{3A} (163.8, SD 70.4) and EQ_{3R} (141.7, SD 54.3). The corresponding p-values, after application of a Bonferroni procedure, are 0.0003 (EQ_1), 0.0000 (EQ_2) and 0.0006 (EQ_3) respectively. Testing the hypothesis that the relation between EQ_{1A} and EQ_{1R}, between EQ_{2A} and EQ_{2R} and between EQ_{3A} and EQ_{3R} can be described by an equation of the form $y = a_0 \cdot x$ resulted in values for r^2 and a_0 of 0.83, 1.72, 0.89, 1.41, 0.92 and 1.19 respectively. The average x-coordinate of the CoP in the three situations with increasing compliance of the supporting surface is 9.1 (SD 5.4), 10.2 (SD 5.4) and 17.1 (SD 6.4). Only the difference between the first and the last ($p = 0.000$) and between the second and the last value ($p = 0.000$) was statistically significant.

Testing the hypothesis that there is a linear relation between $b_{f\text{-}max}$ and EQ_{1R} resulted in values for r^2 and p of 0.27 and 0.009 respectively. Testing the hypothesis that the relation between $b_{f\text{-}max}$ and the product of F_{rg} and EQ_{1R} can be described with an equation of the form $y = a_0 + a_1 \cdot x^2$ resulted in a value for r^2 of 0.54. The average angle of foot tilt, calculated with equation (1), in the three situations with increasing compliance of the supporting surface is 0.014 (SD 0.011), 0.025 (SD 0.016) and 0.032 (SD 0.013). This means that, in these three situations, the standard deviation is respectively 77%, 65% and 43% of the mean value. Testing the hypothesis that the relation between $b_{f\text{-}max}$ and the product of F_{rg} and EQ_{2R} can be described with an equation of the form $y = a_0 + a_1 \cdot x^2$ resulted in a value for r^2 of 0.62. Testing the hypothesis that the relation between $b_{f\text{-}max}$ and the product of F_{rg} and EQ_{3R} can be described with an equation of the form $y = a_0 + a_1 \cdot x^2$ resulted in a value for r^2 of 0.70.

After application of a Bonferroni procedure, the differences between EQ_{1R}, EQ_{2R} and EQ_{3R} are characterized by p-values of 0.0000 (1-2), 0.0000 (1-3) and 0.0001 (2-3) respectively. The average values of EQ_{2A}, EQ_{3A}, EQ_{2R} and EQ_{3R} respectively imply a 38%, 55%, 33% and 45% decrease in EQ relative to the solid surface situation. The average y-coordinate of the CoP in the three situations with increasing compliance of the supporting surface is 27.2 (IQR 28.6), 33.8 (SD 24.3) and 30.1 (SD 19.9). Only the first two situations differed significantly (p=0.02).

Testing the hypothesis that the relation between the pressure under the foot and the decrease in EQ_R as the result of 5 cm and 10 cm foam resulted in values for r^2 and p of 0.01, 0.61, 0.02 and 0.55. Non-linear assessment did not result in higher values of r^2. Testing the hypothesis that the relation between $l_{f\text{-}max}$ and $b_{f\text{-}max}$ can be described with an equation of the form $y = a_0 + a_1 \cdot x$ resulted in a value for r^2 of 0.92. Assessment of the relation between $b_{f\text{-}max}$ and AOS with an equation of the form $y = a_0 + a_1 \cdot x$ resulted in a value for r^2 of 0.96. Assessment of this relation with an equation of the form $y = a_0 + a_1 \cdot x + a_2 \cdot x^2$ did not result in higher values for r^2.

Variable 1 (y)	Variable 2 (x)	Equation	r^2	a_0	a_1
EQ_{1A}	EQ_{1R}	$y=a_0 \cdot x$	0.83	1.72	
EQ_{2A}	EQ_{2R}	$y=a_0 \cdot x$	0.89	1.41	
EQ_{3A}	EQ_{3R}	$y=a_0 \cdot x$	0.92	1.19	
$b_{f\text{-}max}$	EQ_{1R}	$y=a_0 + a_1 \cdot x$	0.27	-300.0	6.4
$b_{f\text{-}max}$	$EQ_{1R} \cdot F_{rg}$	$y=a_0 + a_1 \cdot x^2$	0.54	-1903665	437.0
$b_{f\text{-}max}$	$EQ_{2R} \cdot F_{rg}$	$y=a_0 + a_1 \cdot x^2$	0.62	-1254518	290.0
$b_{f\text{-}max}$	$EQ_{3R} \cdot F_{rg}$	$y=a_0 + a_1 \cdot x^2$	0.70	-749803	194.6
$b_{f\text{-}max}$	AOS	$y=a_0 \cdot x$	0.96	-126.9	3.21

Table 1. The result of linear and non-linear regression analysis on forceplate variables during one-leg stance on three supporting surfaces with varying compliance

DISCUSSION

EQ calculation

There is a clear difference between the values of EQ_A and EQ_R. In all cases, calculation of EQ with the crude amplitudes results in higher values of EQ than calculation of EQ with the slope of the regression line. This corroborates our theory about the effect of foot-tilt independent CoP displacement on the calculation of EQ. Unless the relative contribution of forefoot movement to total CoP displacement is constant, calculation of EQ with the slope of the regression line describing the relation describing the relation between ΔX and ΔM seems preferable. The slope of the regression line between EQ_A and EQ_R can be regarded as an indication of the contribution of forefoot movements to total CoP displacement. When EQ_A and EQ_B would be equal, this slope would be 1.0. The steeper the slope, the larger the contribution of forefoot movement to total CoP displacement. In this view the decreasing slope of the regression line between EQ_A and EQ_R in the order: solid surface, 5 cm foam, 10 cm foam, indicates a decrease in the contribution of forefoot movements to total CoP displacement with increasing compliance of the supporting surface. In the absence of more detailed knowledge about this mechanism for CoP displacement we can only speculate about the reason for this decrease. The most likely cause, in our opinion, is that in situations in which postural control is more arduous, weight is transferred towards the lateral side of the supporting foot. This is supported by the displacement of the average position of the CoP towards the lateral side of the foot observed in this study. Such a shift will result in a decrease of the pressure under the first ray and consequently a decrease in the effect of movements of the first ray on CoP displacement. An alternative cause is that the increased dorsiflexion of the foot on foam affects forefoot movements via altered resting length of the necessary muscles. Whatever its cause, the fact that the amount of foot tilt independent CoP displacement alters with the compliance of the supporting surface, and probably in every situation in which postural control is more arduous, makes EQ_A unsuitable as a measure for the efficiency of foot tilt. This means that the conclusions about effects on EQ based on calculations of EQ_A will have to be re-examined. Because the difference between EQ and EQ_R in terms of percentage is minor, the conclusions about the effect of foam on EQ in the original study[16] remain valid.

Foot breadth

Although the values for p and a_0 (Table 1) suggest that $b_{f\text{-}max}$ and EQ_{1R} increase in parallel, the low value of r^2 for a linear relation between these variables is in contrast with earlier findings [11]. There are several potential causes for this difference. In the first place the number of subjects examined in this study is more then three times (7 v. 24) the number in the cited study. Especially, the present study contains more subjects with large foot breadths. As can be seen in Figure (2) the divergence from linearity as well as the spread of the EQ values occurs predominantly among the wider feet.

In the second place, in equation (2), the relation between $b_{f\text{-}max}$ and EQ_{1R} was shown to be dependent on bodyweight and more of a quadratic nature. As a result of this the likelihood of a quadratic relation between $b_{f\text{-}max}$ and the product of EQ and F_{rg} (Figure 3) was examined. The appropriateness of this approach is shown by the fact that it doubled the value of r^2 (Table 1). Nevertheless, the resulting value of r^2 seems rather low for an adequate description of the relation between foot-breadth and EQ.

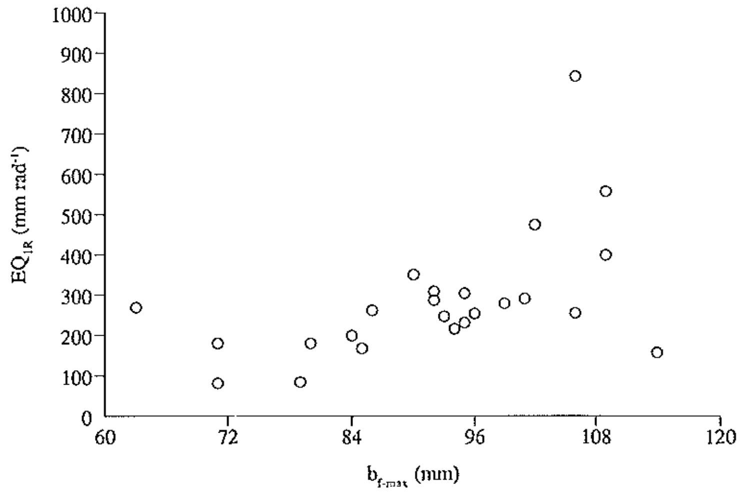

Figure 2. The efficiency quotient of foot tilt on a solid surface (EQ_{IR}) as a function of maximum foot-breadth ($b_{f\text{-}max}$).

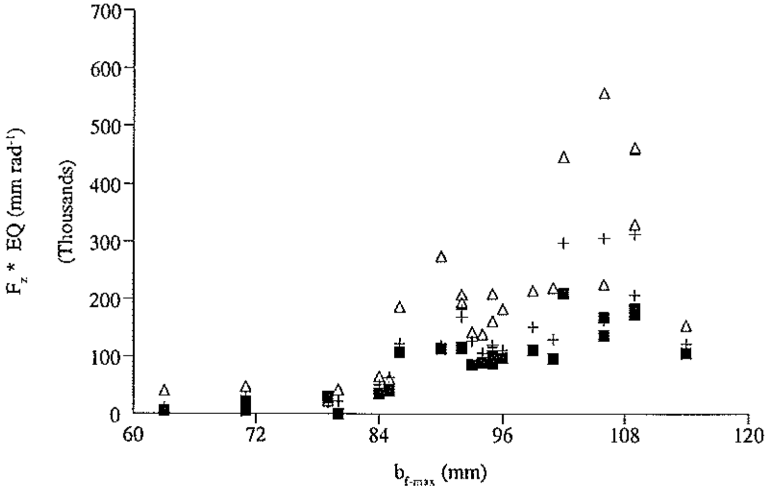

Figure 3. The product of the vertical ground reaction force and the efficiency quotient of foot tilt on a solid surface (Δ), on 5 cm foam (+) and on 10 cm foam (■) as a function of maximum foot-breadth ($b_{f\text{-}max}$).

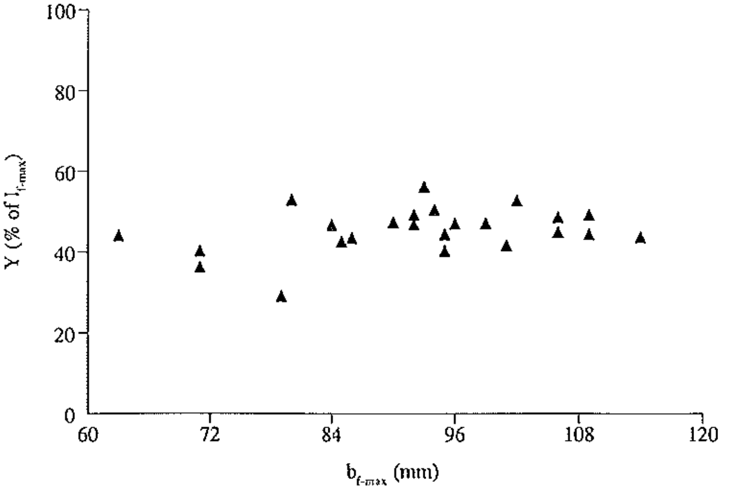

Figure 4. The average y-coordinate of the center of pressure as a percentage of maximum footlength ($l_{f\text{-}max}$) as a function of maximum footbreadth ($b_{f\text{-}max}$).

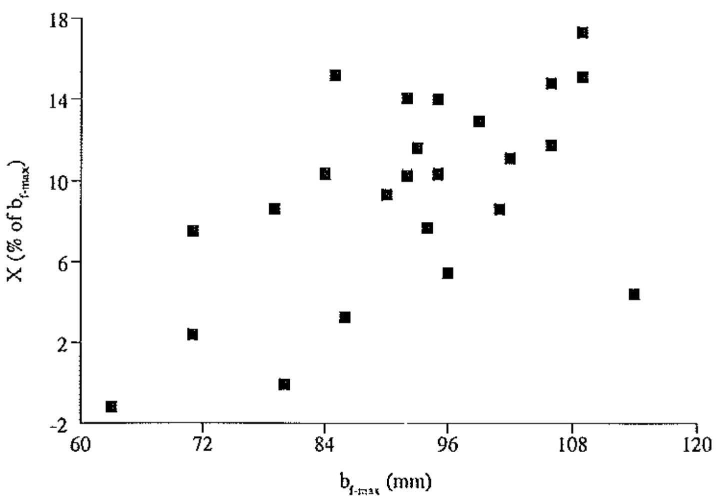

Figure 5. The average x-coordinate of the center of pressure as a percentage of maximum footbreadth ($b_{f\text{-}max}$) as a function of $b_{f\text{-}max}$.

Figure (3) shows that the main deviation from the model still occurs among the wider feet. There are several possible causes for the aberrant behavior of these wide feet. A first explanation is a possible difference in the values of K between the subjects. Besides the compliance of the foam, K probably also depends on the flexibility of the foot. Because currently we have no means to assess the contribution of foot mobility to the value of K, this hypothesis cannot be tested.

A second cause could be a difference in anterior-posterior foot loading that could affect EQ. This is however, not supported by the relatively uniform position of the y-coordinate of the CoP expressed as a percentage of $l_{f\text{-}max}$ (Figure 4).

The third, and to our opinion the most likely cause, is that the value of EQ is affected by the amount of foot tilt. When postural control is more arduous for a subject, increased foot tilt, i.e. increased inversion or eversion of the foot, will shift bodyweight to the rim of the foot. This will narrow the FFB and consequently decrease EQ. Increased inversion with a shift of the bodyweight towards the lateral rim of the foot resulting in a decrease in foot tilt independent CoP displacement and EQ is supported by the increase in the x-coordinate of the CoP, the decrease in the difference between EQ_A and EQ_R and the decrease in EQ_R in the order solid surface, 5 cm foam 10 cm foam. The frontal plane position of the CoP as a percentage of $b_{f\text{-}max}$ can be seen in Figure (5). The fact that the subject with the largest value for $b_{f\text{-}max}$ is an outlier in both figure (3) and (5) also supports this theory. Further support for this theory comes from the amount of foot tilt in the three measurement situations. The standard deviation of this variable as a percentage of its mean value decreases in the order: solid surface, 5 cm foam 10 cm foam. As a result of the more uniform foot tilt the value of r^2 for the relation between $b_{f\text{-}max}$ and EQ_R increases in the order: solid surface, 5 cm foam 10 cm foam. An alternative explanation for this effect is that standing on foam results in such a decrease in K that subject specific differences in K due to differences in foot flexibility become less important. This explanation however, is hypothetical.

Foam thickness

The highly significant differences between EQ_{1R}, EQ_{2R} and EQ_{3R} confirm the hypothesis that a thicker layer of foam causes a stronger decrease in EQ. The 38% decrease in EQ_R when standing on 5 cm foam is comparable with the value found earlier[16]. The 45% decrease in EQ_R when standing on 10 cm foam corroborates our hypothesis of a decreasing effect on EQ of increased foam thickness. The presence of an increase in the y-coordinate of the CoP in the transition from a solid surface to 5 as well as 10 cm foam means that there was an increase in EQ due to an increase in the FFB. As a result of this, the decrease of EQ on 5 as well as 10 cm foam is slightly underestimated.

Foot pressure

The low values for r^2 and p for the relation between p_f and the foam mediated decrease in EQ_R denote the absence of a linear relation between these variables. This is probably caused by the relatively small differences in foot pressures (Figure 6) in combination with a low progressivity of the resistance against compression of the foam. The high value of r^2 for the relation between the measured AOS and $l_{f\text{-max}}$ and $b_{f\text{-max}}$ showed that the former variable can very well be calculated on the basis of the latter two. This method is substantially easier than the method with the laboratory balance.

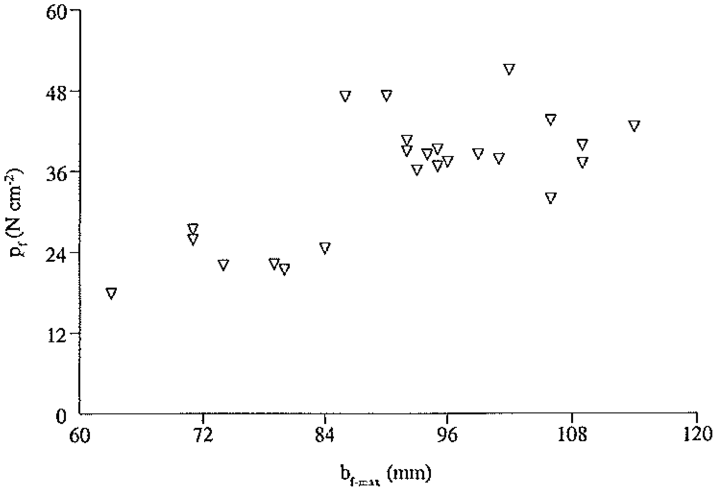

Figure 6. The average pressure under the supporting foot (p_f) as a function of maximum footbreadth ($b_{f\text{-max}}$).

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Chapter VIII

Summary and conclusions

Introduction

The adoption of upright stance is an important (basic) skill. This posture is essential for the performance of many of the activities of daily living (ADL). The sensomotoric integrity of the lower extremities is of great importance for the adoption of upright stance. Despite extensive research on balance control, the relation between frontal plane movements of the lower extremity and balance control was not clear. This thesis attempts to establish a theoretical framework regarding the frontal plane movements of the lower extremities in relation to their function in balance control during one-leg stance.

A much practiced method for balance-research is platform stabilometry. In this method the subject stands on a platform (forceplate) that measures the ground-reaction force in the vertical, fore-aft and mediolateral direction. On the basis of these forces several variables can be calculated that reflect aspects of the movements of the subject on the platform. However, none of these variables was known to be related specifically to the movements of the lower extremity. To correct this omission, a model was developed that describes the relation between the movements of the lower extremity and a forceplate variable namely the point of application of the resultant vertical ground-reaction force. Due to this model it became clear that several factors could affect the operationality of lower extremity movements during postural control in one-leg stance. Knowledge about these factors is indispensable for a correct assessment of the sensomotoric integrity of the lower extremity during postural control.

Summary

In chapter II, due to the dependency of the position of the center of pressure (CoP) on the pressure distribution under the foot, it was hypothesized that, during one-leg stance, there should be a linear relation between foot tilt and the displacement of the CoP in the frontal plane. Because no biomechanical model existed that explains this relation, we presented and validated such a model. Upon validation of this model there appeared to be a close relation ($r^2 = 0.82$) between foot tilt and CoP displacement. It was concluded that, during one-leg stance, frontal plane CoP displacement should not be regarded as a measure of body sway, as is often done, but instead as a measure for the tilting movements of the foot as a mechanism for balance control. It was also shown that foot breadth affects the efficiency of this mechanism, in terms of the amount of CoP displacement per unit of foot tilt (EQ). In view of the importance of CoP displacement for balance control this means that variations in EQ can affect balance control.

In chapter III the effect of the compliance of the supporting surface on EQ is examined. For this purpose a more sophisticated version of the model in chapter II was developed. According to this model an increase in the compliance of the supporting surface will result in a decrease in EQ. The hypothesis was tested that, besides an effect on proprioception, standing on a 5 cm thick sheet of foam affects one-leg stance balance control via a decrease in the amount of CoP displacement per unit of foot tilt (EQ). It was found that standing on foam resulted in a 35% decrease of EQ ($p = .008$), and an increase in amplitude ($p = .010$) and velocity ($p = .012$) of foot tilt. Together this resulted in virtually unaltered values for the amplitude ($p = .995$) and velocity ($p = .613$) of CoP displacement and the medial-lateral ground reaction force ($p = .371$). It was concluded that during one-leg stance on foam, postural control is altered by a decrease in EQ. Young and healthy subjects can fully compensate for this by increasing the amplitude and velocity of foot tilt. There was no measurable effect of decreased proprioception on postural control.

In chapter IV we tested the hypothesis that intra-subject variations in EQ can occur due to positioning bodyweight above the narrow heel or the wide forefoot. Subsequently we examined the effect this had on postural control. Our main outcome measures were the sagittal plane position of the CoP (Y) reflecting foot loading, EQ and medial-lateral ground reaction forces (F_x) reflecting the quality of postural control. It was found that shifting bodyweight towards the rearfoot decreases Y ($p = .002$) as well as EQ ($p = .018$). Shifting bodyweight towards the forefoot increases Y ($p = .000$) and EQ ($p = .064$). In both cases F_x was ($p = .008$ *respectively* $p = .098$) increased and postural control was judged deteriorated. It was concluded that shifting bodyweight to the rear foot or forefoot decreases or increases EQ. Because the changes in EQ encountered here should not deteriorate postural control, other factors are held responsible. The absence or presence of changes in Y can be used to exclude or include the presence of an effect on postural control of EQ or these unknown factors.

In chapter II a relation was established between foot tilt and CoP displacement. In chapter V the hypothesis was tested that, by constraining foot tilt, ankle bracing has a negative effect on postural control via a constraining effect on the amplitude (A_X) or velocity (v_X) of CoP displacement. It was also hypothesized that this effect would increase with increasing values for A_X. It was found that ankle bracing resulted in an increase ($p = .040$) in the average medial-lateral component of the ground reaction force (F_x) reflecting deteriorated postural control. There was also an increase in the average value of A_X and v_X. The relation between A_X and the effect of ankle bracing on v_X can be described by the equation $y = -2.37 + (6.10 \cdot e^{(-x/4.25)})$. This equation means that the constraining effect of ankle bracing on v_X begins at values for A_X of about 4 mm and increases fast. This increase diminishes in the higher amplitude ranges. This seems disadvantageous for healthy subjects who only need bracing near the extremes of the ROM. It could however, be advantageous for subjects with ankle-injury and delayed peroneal reaction time, because it allows more time for ankle bracing by muscle contraction.

In chapter VI the effect of ankle bracing was examined while the subjects stood on a compliant surface (5 cm foam), i.e. in a situation with increased foot tilt. It was hypothesized that in this situation the effects on balance (F_x) and CoP displacement (A_X and v_X) like measured in chapter V would be increased. Compared to the solid surface, ankle bracing while standing on 5 cm foam resulted in a stronger constraining effect on v_X. Nevertheless there was only a minor effect on F_x and A_X. It was hypothesized that this resulted from a shift from foot tilt towards upper body movements as the primary mechanism for postural control.

In chapter VII the efficiency quotient of foot tilt (EQ) was re-examined. EQ can be calculated with the crude amplitudes of CoP displacement (A_X) and ankle displacement (A_M) foot tilt or with the slope of the regression line through A_X as a function of A_M, between these variables. The latter method results in consistently lower values for EQ ($p<0.0001$) that theoretically give a better representation of EQ. The difference between the two methods of calculation reflects the amount of foot tilt independent CoP displacement. The value of EQ is, among others, dependent on the compliance of the supporting surface and decreases in the order solid surface, 5 cm foam, 10 cm foam ($p<0.0001$). The value of EQ is also related to the breadth of the foot. This relation is of the form $y = a_0 + a_1 \cdot x^2$ ($r^2=0.70$). Despite the progressive nature of the resistance of the foam upon compression, an effect of the pressure under the foot on EQ could not be established when standing on 5 cm or 10 cm foam.

Conclusion

During one-leg stance, frontal plane CoP displacement should not be regarded as a measure for total body sway, as is often done, but as a measure for the tilting movements of the supporting foot as a mechanism for postural control. This mechanism was called the foot tilt strategy (FTS). The efficiency of the FTS, i.e., the amount of CoP displacement per unit of foot tilt, is, among others, dependent on the breadth of the foot. The wider the foot, the higher the efficiency. Shifting bodyweight towards the narrow heel or the wide forefoot results in intra-subject variations in the amount of CoP displacement per unit of foot tilt. The more the forefoot is loaded the higher the efficiency of foot tilt. The importance of the FTS is illustrated by the finding that a constraining effect on foot tilt due to ankle bracing results in a decrease in the velocity of CoP displacement and decreased balance during one-leg stance. During unperturbed stance, this effect on the velocity of CoP displacement can only be detected with an amplitude dependent approach. Another negative effect on balance occurs when standing on a compliant surface. This effect is caused by a decrease in FTS efficiency. A combination of ankle bracing and increased compliance of the supporting surface, despite a further decrease in the velocity of CoP displacement, results in a less strong decrease in balance. Calculation of the amount of CoP displacement per unit of foot tilt with the slope of the regression line through the amount of CoP displacement as a function of the amount of ankle displacement, instead of the quotient of these variables, consistently results in lower values for the efficiency of foot tilt. Theoretically these values are a better measure of the efficiency of foot tilt. The difference between the results is a measure for the amount of foot tilt independent CoP displacement.

The knowledge gained from this thesis, is complementary to the existing knowledge about the effects of factors such as body length and weight on postural control, and can be used for modeling the system governing postural control. Additional assessment of the contribution of upper body movements to postural control is a logical next step for this modeling.

Samenvatting en conclusies

Introductie

Het kunnen staan, d.w.z. het handhaven van een verticale lichaamshouding, is een belangrijke (basis) vaardigheid. Deze houding is essentieel voor het uitvoeren van veel van de activiteiten van het dagelijks leven (ADL). Voor het staan is de sensomotoriek van de onderste extremiteit(en) van groot belang. Ondanks intensief onderzoek op het gebied van de balanshanhaving, bleef de relatie tussen de bewegingen van de onderste extremiteit en het handhaven van de balans onduidelijk. In dit proefschrift wordt getracht een theoretisch kader te schetsen waarin verband wordt gelegd tussen de bewegingen van de onderste extremiteit en balanshandhaving in het frontale vlak, tijdens het staan op één been.

Omdat tijdens het staan op twee benen de bewegingen van de benen minimaal zijn, werd besloten de bewegingen van de onderste extremiteit te onderzoeken tijdens de balanshandhaving op één been. Een veelgebruikte methode voor balansonderzoek is platform stabilometrie. Bij deze methode staat een proefpersoon op een krachtenplatform (forceplate) dat de grond-reactiekracht meet in de verticale, voor-achterwaartse en de medio-laterale richting. Op basis van deze krachten kunnen verschillende variabelen worden berekend, die aspecten van de bewegingen van de persoon op het platform weergeven. Van geen van deze variabelen was echter een relatie met de bewegingen van de onderste extremiteit bekend. Om dit probleem op te lossen werd een model ontwikkeld en gevalideerd dat de relatie beschrijft tussen de bewegingen van de onderste extremiteit en één forceplate variabele n.l. het aangrijpingspunt van de resultante, verticale grond-reactiekracht. Met behulp van dit model werd tevens duidelijk dat verschillende factoren van invloed waren op de operationaliteit van de bewegingen van de onderste extremiteit bij de balanshandhaving tijdens het staan op één been. Kennis van deze factoren is onmisbaar voor een juiste beoordeling van de sensomotoriek van de onderste extremiteit(en) tijdens de balanshandhaving.

Samenvatting

In hoofdstuk II werd, aangezien de positie van het CoP afhankelijk is van de drukverdeling onder de voet, gehypothetiseerd dat tijdens het staan op één been er een lineaire relatie zou moeten zijn tussen de kanteling van de voet en de verplaatsing van het CoP in het frontale vlak. Omdat en geen biomechanisch model voor deze relatie bestond werd zo'n model gepresenteerd en gevalideerd. Tijdens validatie bleek er een nauw verband (r^2 = 0.82) te bestaan tussen voetkanteling en CoP verplaatsing. De conclusie was dat, tijdens het staan op één been, de verplaatsing van het CoP in het frontale vlak niet beschouwd moet worden als een maat voor de beweging van het gehele lichaam, zoals vaak wordt gedaan, maar in plaats daarvan als een maat voor de kanteling van de voet als een mechanisme voor balanshandhaving. Er werd ook aangetoond dat de breedte van de voet van invloed is op de efficiency van deze voetbeweging in termen van de hoeveelheid CoP verplaatsing per eenheid voetkanteling (EQ). Gezien het belang van CoP verplaatsing voor de balanshandhaving kunnen variaties in EQ een effect op de balanshandhaving hebben.

In hoofdstuk III wordt het effect van de vervormbaarheid van de ondergrond op EQ onderzocht. Hiervoor wordt een meer geavanceerde versie van het model in hoofdstuk II ontwikkeld. Volgens dit model zou een toename van de vervormbaarheid van het steunvlak resulteren in een vermindering van EQ. De hypothese werd getest dat de balanshandhaving, tijdens het staan op een 5 cm dikke laag schuimrubber, behalve door een effect op de proprioceptie, wordt beïnvloed door via een vermindering van EQ. Het bleek dat het staan op 5 cm schuimrubber (s.g. 40 g dm^{-3}) resulteert in een vermindering van EQ (p = *.008*) met 35%, en een vermeerdering van de amplitude (p = *.010*) en snelheid (p = *.012*) van voetkanteling. Dit resulteert uiteindelijk in vrijwel onveranderde waarden voor de amplitude (p = *.995*) en snelheid (p = *.613*) van de verplaatsing van het CoP en van de grootte van F_x (p = *.371*). De eindconclusie was dat staan op één been op een laag schuimrubber de balanshandhaving voornamelijk veranderd door een vermindering van EQ. Jonge, gezonde personen kunnen dit volledig compenseren door een vergroting van de amplitude en de snelheid van de voetkanteling. Er was geen meetbaar effect van proprioceptie op de balans.

In hoofdstuk IV werd de hypothese getest dat intra-individuele variaties in EQ kunnen ontstaan door het positioneren van het lichaamsgewicht boven de smalle hiel of de brede voorvoet. Vervolgens is het effect hiervan op de balanshandhaving onderzocht. De belangrijkste uitkomstmaten waren de gemiddelde CoP positie in het sagittale vlak (Y) als maat voor de voetbelasting, EQ en de medio laterale, horizontale grond reactiekracht (F_x) als maat voor de kwaliteit van de balanshandhaving. Het bleek dat het verplaatsen van het lichaamsgewicht in de richting van de hiel resulteert in een vermindering van de waarde van Y (p = *.002)* en EQ (p = *.018*). Het verplaatsen van het lichaamsgewicht in de richting van de voorvoet resulteert in een vermeerdering van de waarde van Y (p = *.000*) en EQ (p = *.064*). In beide gevallen bleek F_x toegenomen (p = *.008 respectievelijk* p = *.098*) en de balans verminderd. De conclusie was dat, conform de hypothese, verplaatsing van het lichaamsgewicht naar de hiel of voorvoet resulteert in een vermindering respectievelijk vermeerdering van EQ. Omdat de gemeten veranderingen in EQ geen aanleiding zouden mogen zijn voor de vermindering van de balans, worden andere factoren hiervoor mede verantwoordelijk gehouden. Het aan- of afwezig zijn van een verandering in Y kan worden gebruikt om een effect van EQ of de onbekende factoren op de balanshandhaving aan te tonen dan wel uit te sluiten.

In hoofdstuk II is de relatie gelegd tussen CoP verplaatsing en voetkanteling. In hoofdstuk V is de hypothese getest dat het dragen van een enkelbrace, door een remmend effect op de voetkanteling, de balans verslechtert via een beperkend effect op de amplitude (A_X) of snelheid (v_X) van de verplaatsing van het CoP. We hypothetiseerden tevens dat dit effect zou toenemen bij toenemende A_X. Het bleek dat het dragen van een brace resulteerde in een toename (p= .040) van de gemiddelde medio laterale grond-rectiekracht (F_x), als teken van verminderde balans. Er was ook een toename van de gemiddelde waarde van A_X en v_X. Er was geen amplitude afhankelijk effect op A_X. De relatie tussen A_X en het effect van de brace op de snelheid van het CoP (v_X) kon worden beschreven door de vergelijking $y = -2.37 + (6.10 \cdot e^{(-x/4.25)})$. Dit betekend dat het remmend effect van de brace op v_X begint vanaf A_X waarden vanaf ongeveer 4 mm en daarna snel toeneemt. Deze toename wordt minder sterk in de hoge amplitudes. Dit lijkt nadelig voor gezonden waarbij wie alleen bij extreme bewegingsuitslagen een beperkend effect van de brace nodig is. Het kan echter voordelig zijn bij personen die na enkelletsel een vertraagde reactietijd van de n. peroneus hebben, omdat er meer tijd wordt verkregen voor stabilisatie van de enkel door de aanspanning van de spieren.

In hoofdstuk **VI** wordt het effect van de enkelbrace onderzocht tijdens het staan op 5 cm schuim, d.w.z. in een situatie met versterkte voetkanteling. Er was gehypothetiseerd dat in deze situatie de in hoofdstuk V gemeten effecten op balans (F_x) en CoP verplaatsing (A_X en v_X) zouden zijn versterkt. Het bleek dat, vergeleken met de metingen op een harde ondergrond, het dragen van een brace op een zachte ondergrond alleen resulteerde in een versterking van het effect op v_X. Het effect op F_x en A_X was minder sterk. Dit is waarschijnlijk het resultaat van een verschuiving van voetkanteling naar bewegingen van het bovenlichaam als primair mechanisme voor de balanshandhaving.

In hoofdstuk **VII** wordt de efficiency quotiënt van de voetkanteling (EQ) opnieuw onderzocht. EQ kan worden berekend met de amplitudes van de verplaatsing van het CoP (A_X) en de horizontale enkelverplaatsing (A_M) en op basis van de steilheid van de regressielijn door A_X als functie van A_M. Deze laatste methode resulteert in consistent lagere waarden voor EQ ($p < 0.0001$) welke, op theoretische gronden, een betere representatie zijn van EQ. Het verschil in de twee uitkomsten weerspiegelt de hoeveelheid voetkanteling onafhankelijke CoP verplaatsing. De waarde van EQ is o.a. afhankelijk van de compliantie van het steunvlak en verminderd in de volgorde harde ondergrond, 5 cm schuimrubber, 10 cm schuimrubber ($p < 0.0001$). De waarde van EQ is ook afhankelijk van de voetbreedte. De relatie tussen deze variabelen kan worden beschreven met een vergelijking van de vorm $y = a_0 + a_1 \cdot x^2$ (r^2=0.70). Ondanks de progressief toenemende weerstand van het schuimrubber bij indrukken, kon geen effect aangetoond worden van de druk onder de voet op de vermindering in EQ, tijdens het staan op 5 cm of 10 cm schuimrubber.

Conclusies

Tijdens het staan op één been moeten de verplaatsing van het CoP in het frontale vlak niet beschouwd moet worden als een maat voor de beweging van het gehele lichaam, zoals vaak wordt gedaan, maar in plaats daarvan als een maat voor de kanteling van de voet als een mechanisme voor balanshandhaving. Dit mechanisme noemden wij de foot tilt strategy (FTS). De efficiency van de FTS, d.w.z. de hoeveelheid CoP verplaatsing per eenheid voetkanteling, is o.a. afhankelijk van de voetbreedte. Hoe breder de voet hoe hoger de efficiency. Het verplaatsen van het lichaamsgewicht naar de smalle hiel of de brede voorvoet resulteert in intra-individuele variaties in de hoeveelheid CoP verplaatsing per eenheid voetkanteling. Hoe meer voorvoet belasting, hoe hoger de efficiency van de voetkanteling. Het belang van de FTS voor de balanshandhaving wordt geïllustreerd doordat belemmering van de voetkanteling m.b.v. een enkelbrace resulteert in een vermindering van de snelheid van CoP verplaatsing en een verminderde balans tijdens het staan op één been. Dit effect op de snelheid van de CoP verplaatsing is alleen detecteerbaar bij een amplitude afhankelijke benadering. Een ander negatief effect op de balanshandhaving wordt veroorzaakt door vergroting van de compliantie van het steunvlak. Dit resulteert in een daling van de efficiency van de FTS. Op een meegevende ondergrond resulteert het dragen van een enkelbrace, ondanks een verdere daling van de snelheid van CoP verplaatsing, in een minder sterke vermindering van de balans. Wanneer de hoeveelheid CoP verplaatsing per eenheid voetkanteling wordt berekend met de steilheid van de regressielijn door A_X als functie van A_M resulteert dit in consistent lagere waarden voor deze variabele. Deze waarden zijn, op theoretische gronden, een betere representatie van de efficiency van de voetkanteling. Het verschil tussen de twee uitkomsten weerspiegelt de hoeveelheid voetkanteling onafhankelijke CoP verplaatsing.

De kennis opgedaan in dit onderzoek kan, naast bestaande kennis over de effecten van factoren als lichaamslengte en gewicht op de balanshandhaving, worden gebruikt bij het modelleren van het systeem dat verantwoordelijk is voor de balanshandhaving. Het in kaart brengen van de bijdrage van de bewegingen van het bovenlichaam aan de balanshandhaving is een logische volgende stap voor deze modellering.

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Curriculum vitae

Pieter Hoogvliet werd op 4 juli 1956 geboren in Schoonebeek. In 1973 behaalde hij het MAVO-4 diploma aan de Prof. Casimir Scholengemeenschap te Vlaardingen. De twee daaropvolgende "studiejaren" op de HAVO van deze school werden zonder diploma afgesloten. Hierna werkte hij tot het begin van zijn opkomst in militaire dienst in 1977 als winkelbediende bij de toenmalige Co-oP Nieuwe Waterweg. Tijdens een militaire oefening belandde hij door een val in een (zelfgegraven) schuttersput als patiënt in het medisch circuit. Hier kiemde de idee van een carrière in de gezondheidszorg. Ter verwezenlijking hiervan volgde hij van 1979 tot 1983 het VWO aan de Rotterdamse Avondscholengemeenschap. In deze periode was hij tevens werkzaam als chauffeur bij de Nederlandse Speciaal Drukkerijen te Delft. Van 1983 tot 1991 studeerde hij geneeskunde aan de Erasmus Universiteit te Rotterdam. Van september 1992 tot september 1996 was hij als AIO verbonden aan het Instituut Revalidatie van de Erasmus Universiteit te Rotterdam. In dezelfde periode was hij deeltijds als arts werkzaam op de polikliniek voor handproblematiek (Drs. H.A. Schut) van de afdeling Revalidatie van het AZR Dijkzigt te Rotterdam. Op 1 januari 1997 is hij op deze afdeling begonnen met de opleiding tot revalidatiearts.