Effects of an unstable shoe construction on gait biomechanics and postural control

Master Thesis

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Declaration

I declare that I have developed and written the enclosed Master Thesis completely by myself and have not used sources or means without declaration in the text. Any thoughts from others or literal quotations are clearly marked. This work was not used in the same or in a similar version to achieve an academic grading or is being published elsewhere.

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Preface

The establishment of this master thesis was truly challenging, but thanks to mutual motivation we managed to complete it.

While preparing this thesis, we enjoyed generous support. Therefore, we would like to take this opportunity to thank all those who contributed to the success of this thesis with their professional and personal support.

In particular, our thanks go to our supervisor Dr. Brian Horsak, who helped us with expert advice, when in need. We also thank Jan Swager van Dok for providing the test shoes to all participants. Of course, as the shoes had to be worn as well, we thank our volunteer and very patient participants for taking a lot of time to take the measurements.

Finally, we hope that with this thesis we can provide you with an instructive and pleasant reading.

Lukas Steinbichler & Bernhard Weis

Vienna, 18 January 2019

Abstract

Background: Walking is an integral part of human life and a very complex neuromuscular activity. Postural control and stability are two major factors to ensure natural gait. Shoes are increasingly changing our gait and thus favouring the development of lower extremity pathologies. New shoe constructions have been developed to counteract these changes. The actual effect of these shoes and their benefit for the wearer have been discussed in numerous studies. In this study, a fairly new shoe, the X10D was examined. The main purpose was to investigate the immediate effects of using such a footwear on the lower extremity muscle activity and on postural control.

Methods: 33 participants (19 female / 14 male) volunteered to participate in this study. Surface electromyography data of eight leg muscles were collected while walking barefoot, with regular shoes and with X10D shoes. During quiet standing (double and single leg support), centre of pressure excursion was determined.

Findings: For walking, only slight differences were found in peak muscle activity. When walking with X10D shoes, m. tibialis anterior peak muscle activity was significantly increased (p = 0,002). No significant differences occurred in mean muscle activity. Centre of pressure excursion showed significant differences in anterior-posterior direction when standing on one leg. No significant differences occurred during quiet standing with double leg support.

Interpretation: Results of this study showed that one out of eight muscles reached significance in peak muscle activation and a slight increased instability during standing with single leg support. The increased muscle activity and centre of pressure excursion could have been caused by the special shape of the sole. In order to be able to make more precise statements regarding the effects of X10D shoes further, studies would be necessary.

Keywords: unstable shoes, gait analysis, postural control, electromyography, centre of pressure

Kurzfassung

Hintergrund: Das Gehen ist ein wesentlicher Bestandteil des menschlichen Lebens und eine sehr komplexe neuromuskuläre Aktivität. Die posturale Kontrolle sowie die Stabilität sind zwei Hauptfaktoren für einen natürlichen Gang. Schuhe verändern zunehmend unseren Gang und begünstigen so die Entwicklung von Pathologien der unteren Extremitäten. Um diesen Veränderungen entgegenzuwirken, wurden neue Schuhkonstruktionen entwickelt. Die tatsächliche Wirkung dieser Schuhe und ihr Nutzen für den Träger wurden in zahlreichen Studien diskutiert. In dieser Studie wurde einer dieser Schuhe, der X10D, untersucht. Das Hauptziel bestand darin, die unmittelbaren Auswirkungen der Verwendung solcher Schuhe auf die Muskelaktivität der unteren Extremität und auf die posturale Kontrolle zu untersuchen.

Methoden: 33 Teilnehmer (19 Frauen / 14 Männer) haben sich freiwillig zur Teilnahme an dieser Studie gemeldet. Oberflächenelektromyographiedaten von acht Beinmuskeln wurden beim Barfußgehen, mit normalen Schuhen und mit X10D-Schuhen erhoben. Während des ruhigen Stehens im Zweibein- und im Einbeinstand wurde der Auslenkung des Druckmittelpunkts auf einer Kraftmessplatte bestimmt.

Ergebnisse: Beim Gehen wurden nur geringe Unterschiede in der maximalen Muskelaktivität festgestellt. Beim Gehen mit X10D-Schuhen war die maximale Muskelaktivität des M. Tibialis anterior signifikant erhöht (p = 0,002). Bei der mittleren Muskelaktivität traten keine signifikanten Unterschiede auf. Die Auslenkung des Druckmittelpunkt zeigte signifikante Unterschiede in anterior-posteriorer Richtung, wenn man auf einem Bein stand. Im Zweibeinstand traten keine signifikanten Unterschiede auf.

Interpretation: Die Ergebnisse dieser Studie zeigten, dass einer von acht Musklen, bezogen auf die maximale Muskelaktivität, das Signifikanzniveau erreichte und das sich die Instabilität während dem Einbeinstand leicht erhöht hat. Die erhöhte Muskelaktivität und die stärkeren Auslenkungen des Druckmittelpunkts könnten durch die spezielle Sohlenkonstruktion der Schuhe verursacht worden sein. Um genauere Aussagen zu den Auswirkungen von X10D-Schuhen treffen zu können, wären weitere Untersuchungen erforderlich.

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BW Bernhard Weis
LS Lukas Steinbichler

1 Introduction

Human walking is a very complex neuromuscular activity and postural control as well as stability are two major factors to ensure natural gait (Nigg, Hintzen, & Ferber, 2006). Already at the age of one, children stand on their own two feet for the first time and take their first steps (Orth, 2011, p. 25). The gait pattern is not fully developed at the beginning so the older the child gets and the more it moves, the more secure and economical the gait becomes. However, shoes are increasingly changing our gait and thus favouring the development of lower extremity pathologies (Rao & Joseph, 1992). The sooner children wear shoes, the greater is the likelihood of developing such a pathology, since the arch of the foot forms especially in this early stage of life. To ensure sufficient stability, many shoes are constructed to support the lower extremity muscles and to compensate a possible muscle weakness (Nigg et al., 2006). But these stable shoe constructions come with the disadvantage that the stabilizing musculature they are considering has far less work to do and can thereby atrophy (Landry, Nigg, & Tecante, 2010). Another factor that favours muscular atrophy is the fact that people in our society travel fewer routes by foot and increasingly use other means of transportation. According to the World Health Organization (WHO), adults should walk 10,000 steps a day for a healthy lifestyle. However, we are far from achieving this goal in Austria and also in the other European countries. On average, an Austrian only covers 5351 steps a day (Althoff, Hicks, King, Delp, & Leskovec, 2017).

The stability during walking, however, cannot be achieved only by stable shoe constructions. Another approach to improve stability during locomotion is to strengthen the appropriate lower extremity muscles. In order to counteract the atrophy of the muscles of the lower extremities and the associated diseases, the so-called sensorimotor training has been the method of choice in physiotherapeutic rehabilitation for some years. One way to train these muscles is to exercise on unstable surfaces like a wobble board (Nigg, Hintzen, & Ferber, 2006). Many studies have shown that exercising with unstable surfaces improve the sensorimotor system of the lower extremity (Baldon, Serrão, Scattone Silva, & Piva, 2014). Furthermore, several studies suggest that a standing balance training is beneficial to the prevention of lower extremity musculoskeletal injuries

(Wedderkopp et al., 1999 IN Nigg et al., 2006; Shultz, Silder, Malone, Braun, & Dragoo, 2015).

To make it easier to integrate this training into everyday life and to combine dynamic stability training and locomotion, different unstable shoe constructions have been developed in recent years. The influence of unstable shoe constructions is often considered to modify human gait characteristics positively by strengthening muscles in the human locomotor system and training neuromuscular control (Plom, Strike, & Taylor, 2014). The effects of these shoes on balance, gait and muscle activity have been explored in several studies (Papalia et al., 2015), but the results vary widely and depend on the different groups of subjects. Price, Smith, Graham-Smith, & Jones (2013) also found that the results from shoe model to shoe model are very variable and therefore a generalization is difficult. So, despite those studies, it is still discussed whether these shoes have a benefit by using them and if so, which benefit they have for patients in daily life. Therefore, the use of unstable sole constructions as an adjuvant treatment in the rehabilitative phase should be carefully considered and controlled. For that reason, it is important that a new shoe which pretends to improve the patient’s complaints is established with the support of a scientifically documented study.

Since summer 2013 there is a new shoe construction on the market, named X10D. Through its novel concept, according to the physiotherapist and co-developer Swager van Dok (2015), the shoe should encourage the wearer to increase the pressure on the lateral margin of the foot while walking and thereby obtain a more economical gait pattern. The shoe construction (Figure 1) which lacks the medial part of the sole, causes the shoe to become unstable in the medio-lateral direction. Hypothetically, this instability should be compensated by the wearer. According to Swager van Dok et al. (2015) this should increase muscle activity, improve postural control and as a result, make the gait more economical. In a first study by Swager van Dok et al. (2015), changes in the pressure distribution could already be detected after wearing the X10D shoe for eight weeks. However, the additional benefit this shoe might have in particular, has not yet been scientifically investigated. Therefore, the main purpose of this thesis is to evaluate the influence of this unstable footwear on muscle activity and postural control.

Figure 1 - The X10D shoe model one, which was investigated in this study (Swager van Dok, 2019)

2 Theoretical Background / State of the Art

In the following chapters the physiological human gait, the concept of postural control, the function of lower extremity muscles, basics of electromyography and centre of pressure (CoP) measurements, the sensorimotor training, unstable sole constructions and the concept of the X10D shoe are explained in detail to exemplify the theoretical background of this master thesis.

2.1 The physiological gait

The physiological gait of humans is individually different. The reasons for this are different physical conditions, personal environments, behaviour and the motor and cognitive movement experience (Götz-Neumann, 2016, p.5). The period between two successive contacts of the same foot with the ground is called a gait cycle. There is a differentiation between the gait cycle and the stride length (Figure 2). The stride length indicates the distance between the contact points of the two feet. It begins with the heel contact of one foot and ends with the heel contact of the contra lateral foot. The affiliation of a stride always refers to the leg, which has initial ground contact after completing its swing phase. The term step width is defined by the distance between the two heel centers. While standing, the longitudinal axis of the foot is rotated outwards by about 7° with respect to the locomotion line. This external rotation also persists while walking (Götz-Neumann, 2016, p.5).

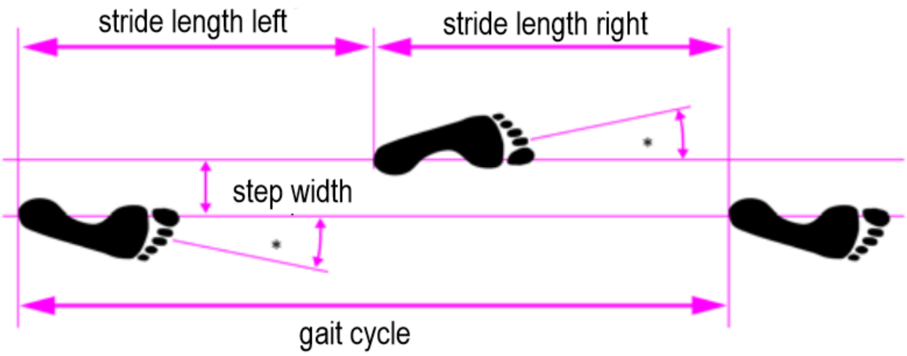


Figure 2 - Graphical representation - gait cycle, stride length, step width

This master thesis uses the Rancho Los Amigos model to describe the gait phases (Gronley & Perry, 1984). The terminology of this model provides a consistent and clear understanding, as shown below in Table 1. In contrast to the conventional description, this model uses neutral terms and can therefore be used for the description of both normal and pathological gait patterns.

Table 1 – Nomenclature according to the Ranchos Los Amigos Model

Gait phases nomenclature

Ranchos Los Amigos	**Traditional**
Initial contact	Heel strike
Loading response	Foot flat
Mid stance	Mid stance
Terminal stance	Heel off
Pre swing	Toe off
Initial swing	Acceleration
Mid swing	Mid swing
Terminal swing	Deceleration

Each gait cycle can be divided into two phases: the stance phase and the swing phase (Figure 3). Götz-Neumann (2016, p. 10) describes the stance phase as the period of the gait cycle in which the foot has contact with the ground. In contrast, in the swing phase the foot is lifted off the ground and brought forward in the air. These two phases can be divided into further sub-phases. In principle, three major functional tasks must be fulfilled during these phases. These tasks are the weight transfer to the foothold, the one leg stance and the forward movement of the swinging leg.

Götz-Neumann (2016, p.11) describes these functions as follows: the weight transfer takes place in the stance phase, more precisely during the initial contact and the loading response. The weight must be taken over to the leg, which has recently been swung forward while the accelerated mass must be broken at the same time. The one leg stance also takes place in the stance phase and will be performed during the mid stance and the terminal stance. In these phases, the body weight must be worn on one leg, while the center of gravity is pushed forward over the leg. Now, the Pre-Swing, Initial Swing, Mid Swing, and Terminal Swing, provide forward swing motion, so the gait cycle can start all over again.

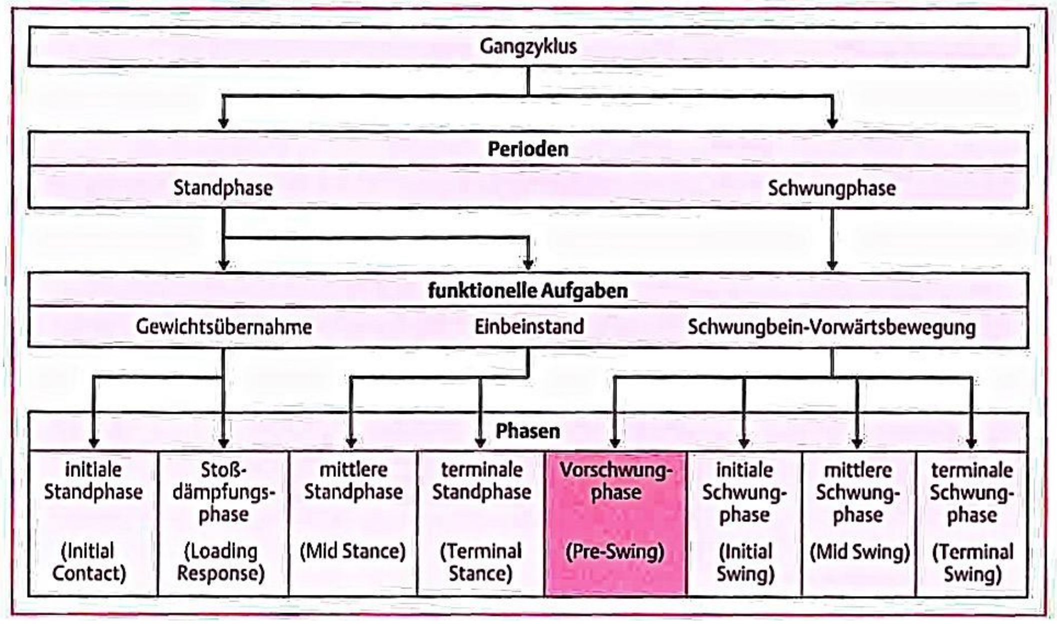


Figure 3 - Division of the gait cycle with the associated periods, tasks and gait phases according to the Ranchos Los Amigos system (Götz-Neumann, 2011, p. 12).

Götz-Neumann (2016, pp. 12-13 & 44-52) describes the gait phases as follows: Initial Contact is the moment when the heel touches the ground and therefore accounts for 0% of the gait cycle. At this time, the hip joint and knee joint of the corresponding leg are in flexion, the upper and lower ankle in zero position or in slight inversion. The Loading Response starts with the heel touching the ground and ends when the opposite leg leaves the ground and the weight of the body is completely absorbed by one leg. The hip flexion does not change, but the knee flexion increases. The upper ankle joint is in a plantarflexion, plus there is an eversion and pronation in the lower ankle joint. In this phase, the landing on the ground is dampened mainly by the knee flexion. In the next phase of the gait cycle, the mid stance, the center of gravity is shifted forward so that it is perpendicular to the forefoot at the end of this phase. Important is the preservation of leg and torso stability. During the mid stance, the hip joint extends to a neutral position, the knee flexion decreases and almost reaches the zero position. The upper ankle joint goes into a dorsal extension, the lower ankle joint remains in eversion, however, the foot is straightened and thus the eversion is reduced. The mid stance is followed by the terminal stance, where the center of gravity is now shifted so far forward that the heel lifts off the ground. The hip extension remains in a neutral position and the knee joint now extends completely. The upper ankle joint and the lower ankle joint are in dorsiflexion, inversion and supination. This phase ends with the initial contact of the contra lateral leg. In rre swing, the hip joint extends more, while the knee joint and the upper ankle joint go into flexion, the lower ankle joint does not change its position. Thus, the toes detach from the ground, so that the foot can now swing forward in the initial swing. At the same time the hip joint flexes and the flexion in the knee joint increases. In the upper ankle joint flexion decreases and the lower ankle joint returns to the zero position. As soon as the swinging leg passes the standing leg and the heels are at the same height, we speak of the mid swing. During the mid swing, the leg is brought forward until the tibia is perpendicular to the ground. Hip flexion increases, knee flexion decreases, and the upper ankle and lower ankle joint return to zero or remain in a neutral position. The terminal swing then completes the gait cycle by barely changing the hip flexion, but now extends the knee, leaving the upper and lower ankle in a neutral position. The toes extend until the foot, or more precisely the heel, is lowered and finally touches the ground again (initial contact) and the gait cycle starts again

2.1.2 Muscle activity during gait phases

For this study the m. gluteus medius, m. gluteus maximus, m. tensor fascia latae, m. quadriceps femoris vastus medialis, m. bicep femoris, m. tibialis anterior, m. gastrocnemius caput medialis, and m. peroneus longus are of importance. Therefore, their anatomical functions and tasks during the gait cycle are described in detail, in the following chapter.

The gluteus medius muscle initiates internal rotation and flexion with its anterior muscle fibers, while the posterior fibers help with external rotation and extension. As an entire muscle it acts as an abductor of the thigh (Platzer, 2009, p. 236). In the gait cycle, it strives to maximize activity during the Loading Response Phase, stabilizing the pelvis and thus the trunk by eccentric muscle activity (Götz-Neumann, 2016, p. 79). It does this with the help of the m. tensor fascia latae and the m. gluteus minimus. These three muscles are also active during the mid stance. During this phase, they lower the pelvis to keep it isometric under the one leg activity. Thereafter, the m. gluteus medius is only active again in the terminal swing, where it prepares for loading response and the associated weight transfer by ensuring pelvic stability (Götz-Neumann, 2016 p. 81, p. 85).The m. gluteus maximus is divided into a superficial and a deep part. Predominantly it acts as an extensor and external rotator in the hip joint and provides a muscular protection against the tipping over of the pelvis to the front (Platzer, 2009, p. 236). During the gait cycle the m. gluteus maximus, especially during the Loading Response phase, actively supports the anterior and posterior thigh musculature to absorb the occurring forces. According to Götz-Neumann, 2016, pp. 90-97, it counteracts the extension in the hip joint and works with its upper fibers against the adduction moment. The m. gluteus maximus becomes active again during the terminal swing to prepare for the new weight takeover.

The m. tensor fasciae latae presses the femoral head against the acetabulum and is also a flexor, internal rotator and abductor in the hip joint. In addition, it supports the front bundles of mm. glutaei medius et minimus (Platzer, 2009, p. 236). When walking, the tensor fasciae latae muscle works mainly during loading response and mid stance, in order to preserve and stabilize the pelvis from slumping along with the other hip abductors. During terminal stance, the muscle takes over the stabilization of the pelvis alone, because the adduction forces are only very low, and the other abductors relax in this gait phase. However, it should also be mentioned that the EMG pattern of m. tensor fascia latae are individually very different (Götz-Neumann, 2016, pp. 90-97).

The m. quadriceps femoris vastus medialis functions as an extensor in the knee joint (Platzer, 2009, p. 248). It is already preparing for loading in the upcoming loading response during the initial contact. There it must provide shock absorption in the knee joint by eccentric muscle activity and therefore reaches its maximum activity in this phase. In the early mid stance phase, the muscle is responsible for the dynamic stabilization of the knee joint but then stops towards the end of the phase (Götz-Neumann, 2016, pp. 68–72). Then, the m. quadriceps vastus medialis is active again in the terminal swing. At this point in the gait cycle the m. quadriceps femoris vastus medialis provides for the complete knee extension needed for initial contact, through concentric muscle work (Götz-Neumann, 2016, p. 76).

The m. biceps femoris consists of a caput longum and the caput breve. In the hip joint, the caput longum works in the sense of a retroversion. The m. biceps femoris bends the knee joint and rotates the lower leg outwards in a bent position (Platzer, 2009, p.250). The m. biceps femoris, like all the other hip extensors, is active while walking in initial contact and loading response. However, it reaches its maximum activity during the terminal swing to slow down the forward movement of the leg. The short head of the m. biceps femoris reaches its maximum activity during the Initial Swing phase, contributing to flexion in the knee joint. Slight activity is also shown in mid swing to control knee extension, if necessary (Götz-Neumann, 2016, pp.80-89 / 90-97).

According to Platzer (2009, p. 258), the anterior tibial muscle causes dorsiflexion in the upper ankle and supination in the lower ankle joint. Götz-Neumann (2016, pp. 54-65) describes the activities of the m. tibialis anterior during the gait phases as follows: During initial contact, the m. tibialis anterior provides pre-positioning of the foot with the remaining muscles of the extensor lodge for the loading response. In the loading response phase, the muscle, along with the m. tibialis posterior, must eccentrically control pronation and stop its activity when full pronation in the lower ankle joint is achieved. In pre swing, the pretibial musculature then prepares for the upcoming dorsal extension in the upper ankle joint. In initial swing, the foot is slowly dorsally extended by concentric muscle work. This dorsiflexion movement continues until the early mid swing phase, when the upper ankle has reached its neutral position and kept held isometrically. Finally, in the terminal swing, the preparation for the weight transfer takes place in the subsequent stance phases.

The m. gastrocnemius is divided into the m. gastrocnemius caput medialis and the m. gastrocnemius caput lateralis. Together with the m. soleus they form the m. triceps surae (Platzer, 2009, p. 262). According to Platzer, it is the muscle of plantarflexion par excellence. This muscle is able to lift the weight of the body when standing and walking. Therefore, the gastrocnemius muscle is particularly important when walking, as it is effective not only when lifting the heel, but also when bending the knee (Platzer, 2009, p. 262). During the gait cycle, the calf muscles are predominantly active in mid stance and terminal stance, thus controlling the forward movement of the tibia and stabilizing the knee joint. Simultaneously, the activity of the m. triceps surae supports lifting the heel (Götz-Neumann, 2016, pp. 67-72 / 80-89).

The m. peroneus longus, together with the m. peroneus brevis, is the strongest pronator in the lower ankle joint (Platzer, 2009, p. 260) and also acts as a plantar flexor in the upper ankle joint. It acts predominantly during the one leg stance in the gait cycle, in the mid stance, as well as in the terminal stance. Especially in the terminal stance, the m. peroneus longus achieves maximum activity together with the plantar flexors of the upper ankle joint. In this phase, together with the m. peroneus brevis, it is especially responsible for the lateral stability of the lower ankle (Götz-Neumann, 2016, p. 65).

2.2 Basics of electromyography

Electromyography (EMG) is defined by Banzer, Pfeifer, & Vogt (2004) as a method for determining excitation and contraction states of skeletal muscle. According to Konrad (2005, p.7) the EMG signal arises from the depolarization-repolarization process of the action potential, shown in Figure 4. The prerequisite for this is the rest potential, which is maintained by a constant ion exchange (ion pump) through the semipermeable membrane of the muscle cells. This results in a negative charge of about -80 mV of the cell interior in the non-contracted state of the muscle. If an impulse from the central nervous system is passed on to the motor end plate, transmitter substances are released, which temporarily change the diffusion properties of the muscle membrane and allow sodium to flow increasingly into the cell interior. There is a short-term depolarization of the muscle fiber membrane and a charge change to about +30 mV, which is, however, immediately restored by an active compensatory ion backflow.

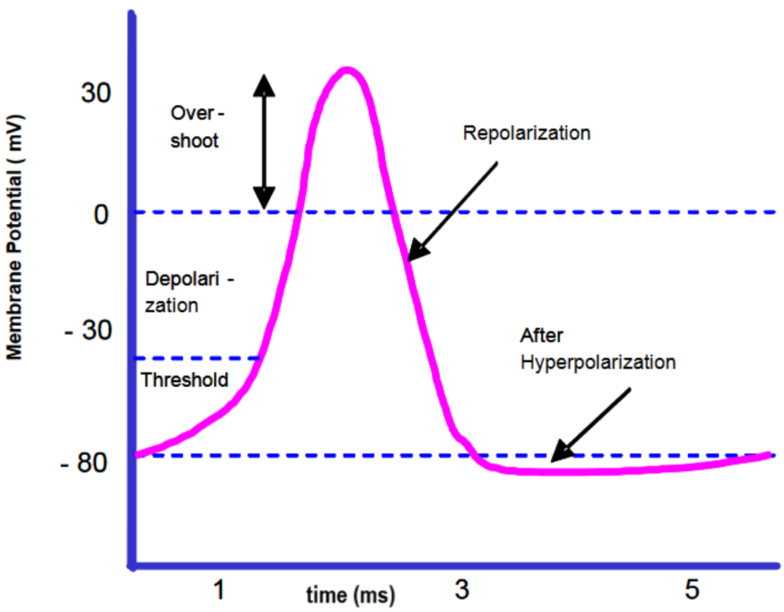


Figure 4 - Graphical representation of a depolarization-repolarization process (Konrad, 2005, p.7)

Konrad (2005), describes that the repolarization phase is followed by a hyperpolarization phase (Figure 4), in which the cell is not excitable for a short time and that the action potential, starting from the motor endplate, continues bidirectionally along the muscle fiber causing the muscle to contract.

In principle, a distinction can be made between two different methods, namely the clinical EMG and kinesiological EMG. According to Banzer et al. (2004, pp. 166-167) the clinical EMG is used for the diagnosis of neuro- and myopathies. Nerve conduction velocity, discharge rates of motor units as well as the duration, amplitude and shape of individual action potentials are investigated with needle or wire electrodes. With the kinesiological EMG, the focus is on the investigation of relationships between muscle actions, movements and forces. In most cases, surface electrodes, in some cases also needle electrodes are used.

The electrodes detect the potential differences between the individual motor units, which are superposed and represented graphically in an interference pattern (Banzer et al., 2004).The actual measurement signal is the raw EMG, the superposition of the interference pattern of all single measured muscle fibers as shown in Figure 5 below.

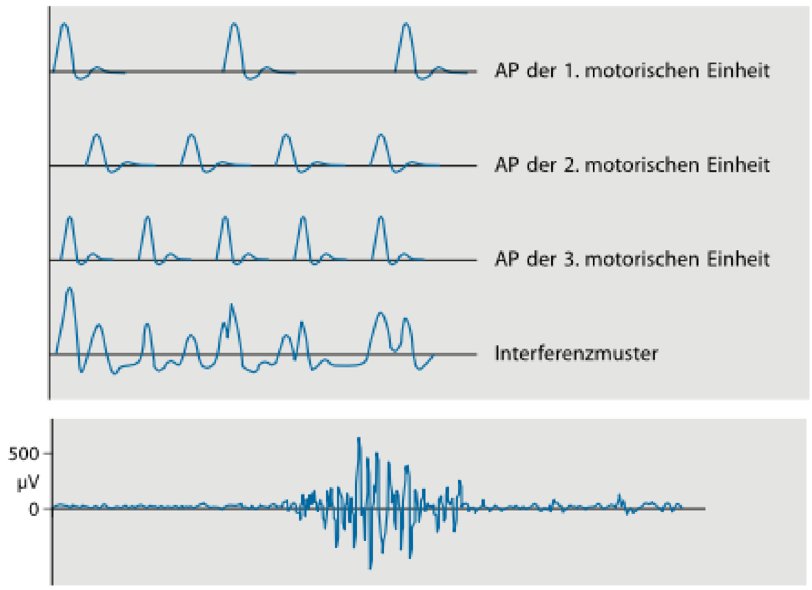


Figure 5 - Action potentials and interference patterns (top), raw EMG (bottom) (Banzer et al., 2004, p. 168)

On the way from the muscle fiber to the electrode, the signal may be affected due to different factors. Konrad (2005) describes five groupings in which the confounding factors can be classified:

1. Tissue properties

The conductivity of the tissue varies depending on tissue type, thickness and temperature. These characteristics could differ between humans as well as individual derivation points, due to what a direct comparison between different discharge points is not valid.

2. Physiological Cross Talk

A significant portion of the measured signal can also be produced by neighbouring muscles.

3. Distance change between muscles and electrodes

The EMG signal amplitude is changed by any distance change between the muscle and the electrode. This is above all a problem in dynamic motion studies.

4. External interference voltage

Interference voltages can be caused by poor or ungrounded external devices, which are noticeable as artefacts in the measurement signal.

5. electrodes and amplifiers

The quality of the electrodes and the internal amplifier noise can influence the measurement results as well.

By appropriate preparation, especially a correctly performed skin preparation and electrode placement, as well as by controlling the given conditions, most influencing factors can be minimized. With an EMG amplifier, noise can be suppressed to reduce artefacts (Figure 6) by measuring the potential difference between the electrodes and thereby eliminating external interference (Konrad, 2005).

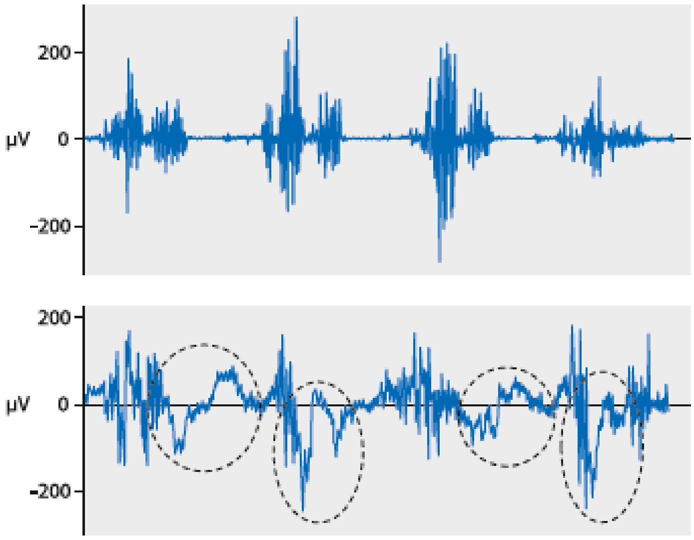


Figure 6 - EMG measurement without artefacts (top) and with artefacts (bottom). The artefacts were marked with circles in the lower picture (Banzer et al., 2004, p. 171).

The raw EMG signals can be evaluated and interpreted in different ways. Banzer et al., 2004 describe three evaluation methods:

1. time-related evaluation
2. amplitude-related evaluation and
3. frequency analysis

The time-based evaluation determines the beginning and the end of muscle activity. An associated problem is the distinction between low voluntary muscle activity and background noise, which are difficult to distinguish because of the stochastic nature of EMG (Banzer et al., 2004, p. 172). In the amplitude-related evaluation, Konrad (2005) names three standard amplitude parameters, which are shown in Figure 7. The EMG peak is the value that describes the maximum amplitude swing. However, this should be interpreted with caution, since it is very variable. The amplitude mean is not really sensitive to small time differences in contractions. According to Konrad (2005) it is one of the most important parameters, since it best describes which neuromuscular gross input has contributed to a muscle movement and is therefore very well suited for comparative analysis. The integral is the area under the EMG curve that is directly dependent on the duration of the contraction. To calculate the minimum, maximum, integral and mean, the raw EMG signal must first be full wave rectified by positivizing all negative excursions by mathematical magnitude (Konrad, 2005).

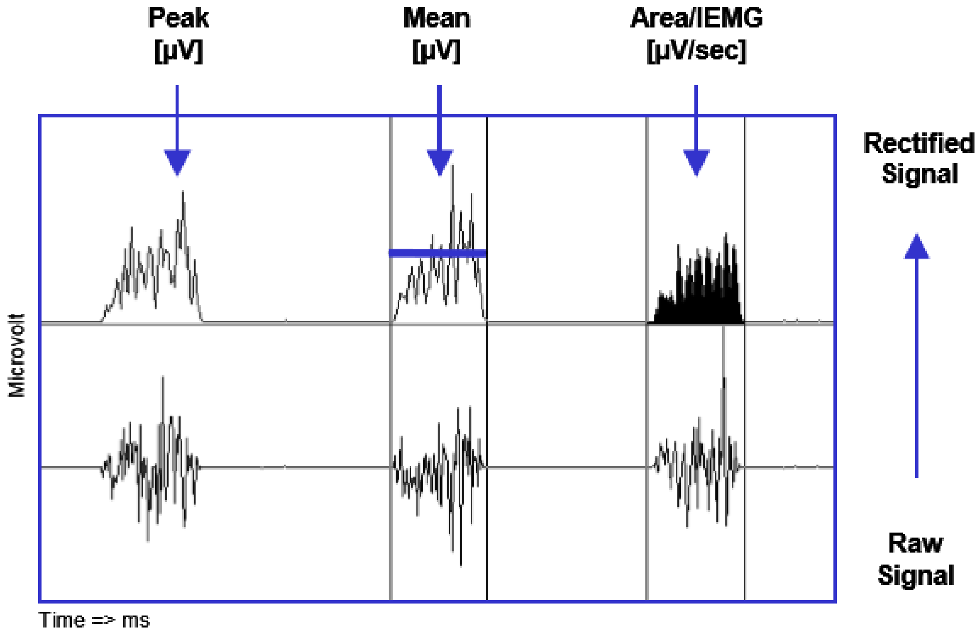


Figure 7: EMG standard amplitude parameter based on the corrected EMG curve (Konrad, 2005, p. 26)

The frequency analysis is mainly used in the study of muscle fatigue (Banzer et al., 2004, p. 175). Again, Konrad (2005) describes three parameters for describing the total frequency power spectrum. Total power is the area under the curve divided by the median frequency in the middle. The mean frequency, which corresponds to the mathematical mean of the spectrum curve, together with the median frequency are the two most important parameters in the frequency analysis of static fatigue contractions (Figure 8).

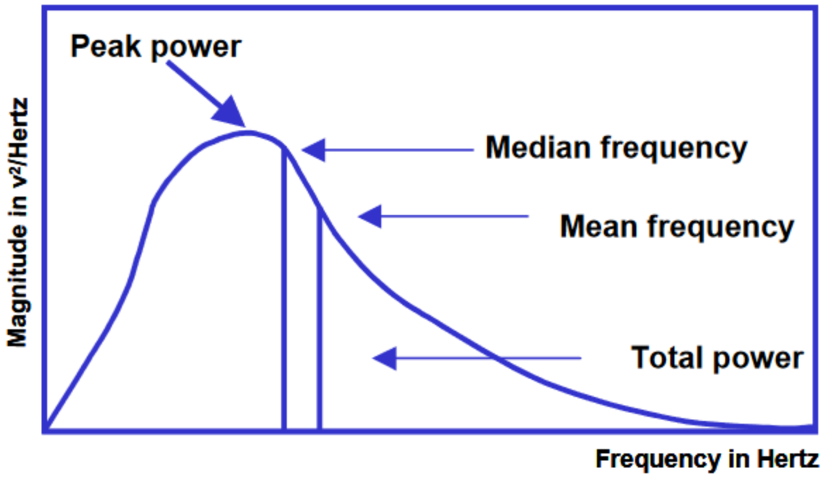


Figure 8 - Schematic representation of the power spectrum (Konrad, 2005)

Another problem of the EMG analysis is that the amplitude values, in addition to the confounding factors already described, may depend on the subject's daily condition. One way to eliminate this variability is to normalize the amplitude value to a reference value. In the Maximum Voluntary Contraction concept (MVC concept), the reference value used is a previously performed maximum contraction of the measured muscle. The amplitude data is then expressed as a percentage of the reference value of the maximum contraction to eliminate the influence of local discharge conditions (Konrad, 2005). A great advantage of MVC normalization is that the values normalized give a better understanding of the actual innervation and effort of the target muscle and how demanding certain activities for the muscle being measured are. Likewise, EMG data can be compared quantitatively between individuals due to the elimination of derivation conditions by MVC normalization (Konrad, 2005). According to Konrad (2005) disadvantages of MVC normalization are, that the amplitude normalization can only be performed with healthy volunteers and additionally that not all trained persons manage to contract a muscle to the maximum. Static muscle testing can be considered problematic in the extent that the muscle length in these tests may differ from the muscle length in the dynamic motions under investigation and therefore the tests are not representative.

2.3 Postural control

Life developed in the presence of gravity and it has always been recognized that the posture is obtained through tonic muscle contractions that act against gravity and stabilize the positions of body segments (Ivanenko & Gurfinkel, 2018).

Posture control research is shaped by many concepts. In principle, two different levels are distinguished in the postural control system. One level determines the distribution of tonic muscle activity, „posture" and the other level is used to balance internal or external disturbances, “equilibrium”. Although these two levels are intrinsically linked, neurophysiological as well as functional considerations point to different neuromuscular bases. Skeletal muscle and its unique structure and properties should also be considered in order to understand peripheral factors affecting posture regulation. Subsequently, various concepts and the neuromechanical principles of posture are developed (Ivanenko & Gurfinkel, 2018).

A simple scheme for illustrating upright posture is based on the idea of the inverted pendulum and the presence of center of pressure (CoP) oscillations as an important measure of positional stability. In this simplified model the center of body mass (CoM) is the only controlled variable (Winter, Patla, Ishac, & Gage, 2003).

Ivanenko & Gurfinkel (2018) claim that in a steady state the CoP swings on both sides of the CoM to keep it in a constant position between the two feet, shown in Figure 9 below. Since the body center (CoM) is relatively high in the trunk, the posture is inherently unstable. However, the variations in the CoP are not dependent on the height of the CoM. For example, Figure 8 shows typical examples of the center of pressure fluctuations in calm standing in cats, dogs and humans. Note the similar CoP oscillations (~1-2 cm) beyond the distances in the middle of the body. Therefore, the simple principle "the lower the CoM, the smaller the CoP oscillations" is deceptive or cannot be generalized to animals of different sizes. In addition, the amplitude of the CoP oscillations is much smaller than the actual support base and would likely provide stability, even if it were larger.

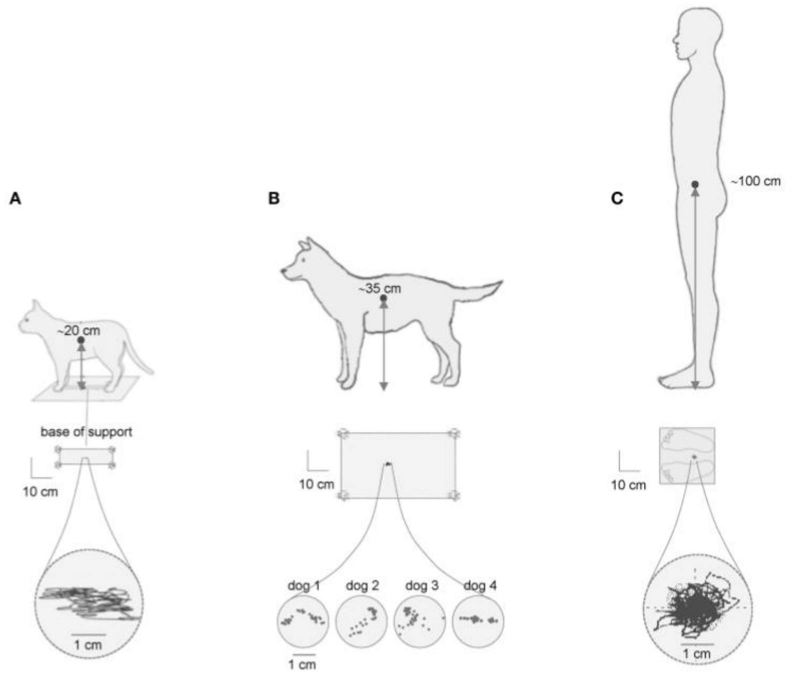


Figure 9 - Center of pressure (CoP) fluctuations during quiet standing in the cat (A), dog (B) and human (C). Note comparable CoP oscillations (~2 cm) in quadrupeds with regard to human despite the 5-fold difference in the height of the center of body mass over the support (Ivanenko & Gurfinkel, 2018).

It is therefore important to emphasize that simple biomechanical considerations can only conditionally explain postural behaviour. The upright posture in humans is provided in part by passive structures such as bones, joints, ligaments and muscles, but it also requires active contraction in the lower extremities, trunk and neck (Ivanenko & Gurfinkel, 2018). Postural tone control is not simple and requires special neural circuits. It requires detailed information about the underlying neural circuits and the underlying cellular processes involved in producing longer muscle force and stiffness (Ivanenko & Gurfinkel, 2018). Multiple sensory and motor areas have evolved and integrated throughout our body's life history to allow accurate regulation of body orientation in the gravitational field.

2.3.1 Structural and functional complexity of postural muscles

The structure and function of the skeletal muscle allows a wide range of activities, from fast or powerful movements to the permanent maintenance of the body relative to gravity. Postural tone is generally considered to be low muscle tension, which is observed in both the distal limb muscles and in the proximal musculature of the trunk and neck. Nevertheless, one cannot reflect the postural tone by exclusively considering the neural input of the subcortical and cortical structures. Recent findings from biochemical and biomechanical research have forced a reassessment of structural and functional muscle complexity (Knight, 2016).

Ivanenko & Gurfinkel (2018) explain that the sliding filament theory for muscle contraction has been extended to include regulatory and cytoskeletal proteins responsible for the viscoelastic properties of muscle and the efficiency of force production, key control of stimulation regulation. In the context of postural function of skeletal muscle, the elastic properties of skeletal muscle and muscle tension are closely related to regulatory and cytoskeletal proteins. Although the posture musculature is rather small, it is important to emphasize that neither posture is passive. The low activities of the neck, trunk and extremity muscles determine the resting tension, as well as the axial tone, the individual postures, the facial expression and others (Caneiro et al., 2010; Wright, Gurfinkel, Nutt, Horak, & Cordo, 2007). Long lasting maintenance of postural muscle activity for hours is associated with low energy costs. Postural activity is usually achieved by slow muscle fibers that are more resistant to fatigue. In addition to the selective activation of corresponding muscle fibers, a little-understood but fascinating aspect of postural muscle tone, includes the mechanisms of muscle elasticity, strength enhancement and energy conservation (Ivanenko & Gurfinkel, 2018).

Postural fluctuations are still poorly understood, although they are among the best studied mechanisms. On the one hand, approaches and controversies for their description, but on the other hand, the postural control as a complex and dynamic system process of interacting components can only be understood if a sufficient description of the underlying processes is touched on. It is discussed whether the body center of gravity fluctuations are an image of imperfection, a systemic white noise in the sense of a purely stochastic process or a sophisticated variant of the physiological system, to create equilibrium in a labile environment of serially and parallely coupled, seemingly redundant joints by allowing some degrees of flexibility.

Ivanenko & Gurfinkel (2018) note that the upright bipedal stance is traditionally described to depend on sensory (vision, vestibular, and somatosensory) input to provide postural equilibrium and a proper alignment of body segments with respect to gravity. The nature of multisensory interactions has been the subject of plenty of studies. From the conceptual viewpoint we will consider below the three common theories of postural regulation that have been rather influential in many experimental studies and mathematical models of human posture control:

1. the posture control system is linear
2. the posture control is determined by reflexes
3. the posture control is equilibrium control

2.3.1.1 Non-linear Properties of the Posture Control System

Small movements accompany the maintenance of any posture. Typically, unless human posture is unstable, body segment oscillations do not exceed 1-2° of joint movements and the CoP oscillations are about 1-2 cm. The fact that postural oscillations are small, supports the assumption that the system is linear within a limited range of movements and, therefore, linear computational models and analyses can be applied (Ivanenko & Gurfinkel, 2018). While this assumption is valid to some extent and many studies provided important information about postural strategies, one should keep in mind that there is also substantial non-linearity in the postural control system, which is often overlooked.

First of all Ivanenko & Gurfinkel (2018) claim that, some non-linearity exists already at the level of muscles, since their resistance to small angular perturbations is much higher than the resistance to larger perturbations. Even though the short range stiffness of active calf muscles might not be sufficient to fully compensate the body sway during quiet standing, its contribution is definitely essential (Gurfinkel et al., 1995 IN Ivanenko & Gurfinkel, 2018). Ivanenko & Gurfinkel (2018) sum up, that by ignoring the non-linear dependence of ankle stiffness on sway size may lead to serious misinterpretation of the results of experiments that use mechanical perturbations or sensory manipulations such as eye closure, movable or unstable support surfaces and sway-referencing (Loram, Maganaris, & Lakie, 2007). Secondly as postural oscillations are small, there is considerable nonlinear redistribution of the internal shifts of muscle fibers, tendons, and soft tissues in the body. Due to the compliant achilles tendons, there is a paradoxical shortening of calf muscles when the body sways forward and a lengthening when the body returns. As a result, the postural role of the numerous calf muscle spindles in the detection of body sway remain uncertain (Loram, Maganaris, & Lakie, 2004). In addition, the control of balance and internal shifts of muscle fibers, ligaments and soft tissues is not limited to distal joints. Furthermore, constant axial muscle activity is required to stabilize the trunk and head and if necessary balance movements of the distal body parts to maintain postural stability (Ivanenko & Gurfinkel, 2018). Furthermore, the human foot is significantly deformed while standing upright due to small CoM shifts and deformations of the soft tissues and the arch of the foot. The postural activity of numerous foot muscles further contribute to the plasticity of the human foot. However, Ivanenko & Gurfinkel (2018) note, that many postural studies like those of Gatev, Thomas, Kepple, & Hallett, (1999) and Winter, Patla, Ishac, & Gage (2003) tend to focus on the simple hinge action of the ankle joint.

According to Ivanenko & Gurfinkel (2018), the shift of paradigms in future studies may be related to the development of non-linear approaches, although complexity of the model may come at the cost of understanding. Therefore, it is necessary to find a compromise between the usage of linear approaches and more complex postural models. Nevertheless, even if we apply linear computations due to their simplicity, we need to keep the nonlinearity in the neuromuscular control of posture in mind.

2.3.1.2 Posture Control as a Summation of Postural Reflexes

Ivanenko & Gurfinkel (2018) note that early postural control studies emphasized that postural mechanisms were based on reflexes and made some important findings. The idea that stretch reflexes and sensory feedback in connection with the concept of servoregulation was very influential on earlier models and investigations of postural control.

However, it has been concluded that the postural reflexes alone do not provide a complete explanation for the complexity of postural control, which includes not only reflexes but also anticipatory adjustments, contextual sensorimotor adaptations, a postural body scheme, and the integration of posture and movement (Massion, 1994 IN Ivanenko & Gurfinkel, 2018).

Ivanenko, Grasso, & Lacquaniti (1999) claim, changes in the line of vision may also modulate postural responses, consistent with supraspinal or cognitive influences on postural control. This is probably because gaze is an important frame of reference for the internal model of spatial orientation. Since automatic postural reactions are made in consensus with the internal representation of the body scheme, one can conclude that it not only serves to consciously perceive the position but is also the basis for the planning and implementation of the motor activity (Ivanenko & Gurfinkel, 2018).

According to Ivanenko & Gurfinkel (2018), a complex interaction of physiological mechanisms, high level processing of sensory information in accordance with the postural body scheme and on the individual's expectations, goals, cognitive factors and prior experience are responsible to control the balance during both standing and movement. The notion of body scheme has received attention in a large context of contemporary motor control to understand adaptability of reflex modulation, processes like the state estimation, prediction and learning, and to bridge the gap between cognitive and motor functions (Ivanenko & Gurfinkel, 2018). In sum, postural control is no longer considered a system or a given set of equilibrium reflexes but rather a motor skill.

2.3.1.3 Posture Control and Equilibrium Control

Many articles on postural control typically claim that sensory information from somatosensory, vestibular, and visual systems is integrated to maintain equilibrium. Therefore, a consistent piece of research focusing on the postural equilibrium examines how sensory inputs are rebalanced or how neural strategies change in different situations to avert disorders and control balance. However, Ivanenko & Gurfinkel (2018) claim, that the posture control system has to cope with both tasks at the same time. One sets the distribution of tonic muscle activity ("posture") and the other is assigned to compensate for internal or external disturbances ("equilibrium") and question if these two tasks are equivalent.

At the beginning it should be said, that in controlling movement and maintaining a firm posture, different neural circuits are used in the brain stem, cerebellum, motor cortex and hippocampus (Shadmehr, 2017).

Shadmehr (2017) suggested that the need for a "holding circuit" may result from the need to maintain a constant "sensory state" while circuits responsible for the movement of the body part change its sensory state. Since the two tasks, motion and standing still are inherently linked, there are also overlaps and interactions between these circuits. Nevertheless, they differ considerably. Neurophysiological data on various modalities of control of gaze, head movements, arm movements, posture and locomotion indicate that different interneurons and motor neurons show activity bumps during transient movements versus a sustained discharge level during posture (Shadmehr, 2017). Hess (1954 IN Ivanenko & Gurfinkel, 2018) says that a similar concept can be applied to the control of phasic and tonic postural muscle activity. As far as the postural tone is concerned, it emerges from several supraspinal centers, including the reticular formation, the vestibular nuclei, the cerebellum and the mesodiencephalic nuclei (Hess, 1954 IN Ivanenko & Gurfinkel, 2018). These brain regions may have sustained long-lasting activity that allows prolonged excitement and inhibition of executive motor systems. In addition, there are specialized spinal cord paths and a specialized core muscle activation during various postural and motor tasks. Slow and fast processes in the central nervous system are also often related to the control of muscle tone and phasic muscle activity (Ivanenko & Gurfinkel, 2018).

Ivanenko & Gurfinkel (2018) claim, that in addition to the operative control provided to compensate for deviations from a reference position, the postural control system includes at least one additional level, which elaborates this postural "set" taking into account the energy cost of standing, the position of body segments, the muscle torques and the demands for stability and security. Ivanenko & Gurfinkel (2018) consider posture and equilibrium to be conveyed by distinct neural circuits, while posture-stabilizing mechanisms may be responsible for the control of equilibrium relative to the superiorly determined postural set.

To sum up, the central nervous system is able to combine mobility with stability and the nature of posture-movement interactions is a well-known problem in neuroscience. Sherrington (1906) very aptly described the latter aspect as early as in 1906, "posture follows motion like a shadow." Movements are even anticipated. Tonic muscle activity and postural control require special neural circuits. Appropriate postural tone is an integral part of any movement and muscle tone disorders can in turn affect the performance of movements. We need to further explore the neuromuscular underpinnings of postural tone and how it is generated and maintained, to better understand control of posture and movements (Ivanenko & Gurfinkel, 2018).

2.3.2 Postural stability and CoP measurements

Postural stability is an important component for maintaining an upright posture and for maintaining balance in activities and movements of everyday life. Postural stability plays an important role, especially for the elderly people as the risk of falling and misalignment is usually increased in older age groups as a result of a progressive balance disorder. Also in sports, balance problems can lead to serious injuries (McGuine IN Ruhe, Fejer, & Walker, 2010). Postural stability, therefore, has important implications for sports and rehabilitation.

Various methods are used today for assessing postural stability. Frequently, an evaluation of parameters describing CoP deviations is used to measure the postural stability and make it comparable. These measurements are possible because the CoP signal is proportional to the ankle torque. The ankle torque describes a combination of descending motor commands as well as the mechanical properties of the surrounding musculature (Baratto, Morasso, Re, & Spada, 2002). The most commonly used measurement parameters are related to judicious measurements such as pendulum distance, velocity and traversed areas and are based on successive positions of the CoP in the plane of the force platform (Ruhe et al., 2010).

Many factors contributing to postural control have been identified. This postural control system depends on the unimpaired ability to correctly perceive the environment through peripheral sensory systems, as well as to process and integrate vestibular, visual and proprioceptive inputs at the central nervous system (CNS) level. The CNS employs different strategies to form appropriate muscle synergies needed to maintain equilibrium, depending on whether the task at hand is static or dynamic (Balasubramaniam & Wing, 2003).

In addition to individual perceptual and motor skills, musculoskeletal characteristics and task constraints, as well as the area of support in terms of foot position play an important role in postural stability. Balasubramaniam & Wing (2003) claim that the measurement methods of the human standing position can be roughly divided into the following three main groups:

1. Displacement of the body segment during the standing position
2. Muscle activity to maintain postural balance
3. Measurement of the motion and the patterns of the center of mass (CoM) or center of pressure (CoP).

According to Winter (1995), body segment displacement refers to the change in position of body segments such as head or trunk during adaptive movements in order to maintain balance.

While controlling balance, the muscle action appears to be an anticipatory feed-forward mechanism that is determined by an internal model of the inverted pendulum and acts in the long term. It is designed to prevent the body from falling and to align the body around its reference point (Baratto et al., 2002). In contrast, the intrinsic feedback due to mechanical properties of ankle muscles operates with a zero delay in the short-term in order to slow down the fall of the inverted pendulum. The inverted pendulum model connects the controlled variable (CoM) with the controlling variable (CoP) (Gatev et al., 1999). The complementation of this mechanism by the feed-forward control is necessary, because the muscle stiffness itself is not sufficient to stabilize the body when it comes to a critical level of displacement (Baratto et al., 2002).

CoP can be defined as the position of the global ground reaction force vector that accommodates the sway of the body. In simple terms, it is the point at which the pressure of the body would be if it was concentrated in one spot on the ground. However, this measure is not a correct record of the body sway but a measure of the activity of the motor system in moving the CoP. The point equivalent of the total body mass in the global reference system is called center of mass (CoM) and is commonly accepted to lie around the S2 vertebral level in normal upright posture (Gard, Miff, & Kuo, 2004). The relationship between CoP and CoM during stance, where CoP oscillates on either side of the CoM were demonstrated by Lafond, Duarte, & Prince (2004). While CoP theoretically completely coincides with CoM at low sway frequencies below 1 Hz (Winter, 1995), its displacement during sway always exceeds that of the CoM (Lafond et al., 2004).

Out of these, one of the most commonly used tools to investigate the complex balance system of postural control is the stabilogram. The stabilogram is a measure of the time behaviour of the CoP of a person standing on top of a force platform consisting of a rigid plate supported by force transducers. Postural sway monitored in quiet standing represents the integrated output from the complex interaction between the balance systems mentioned above. As the understanding of these balance mechanisms has evolved in recent decades, there is also a major change in the literature regarding study design and instruments used to investigate CoP. Ruhe et al. (2010) concluded, that even though the evaluation of CoP excursions is a commonly used method for measuring postural stability no standardization of this method exists.

2.3.3 CoP measurements via force plates

In principle, the measurement always follows the following basis. The idea is to obtain an estimator for the body center of gravity via force transducers, which are simply arranged rectangularly in the apparatus. If we describe the measured forces of the sensors and the coordinates of the sensor locations relative to the specified origin, the coordinates of the vertical projection of the CoM of a rigid object onto the support surface can be calculated (named center of mass, CoM) (Borg, 2005).

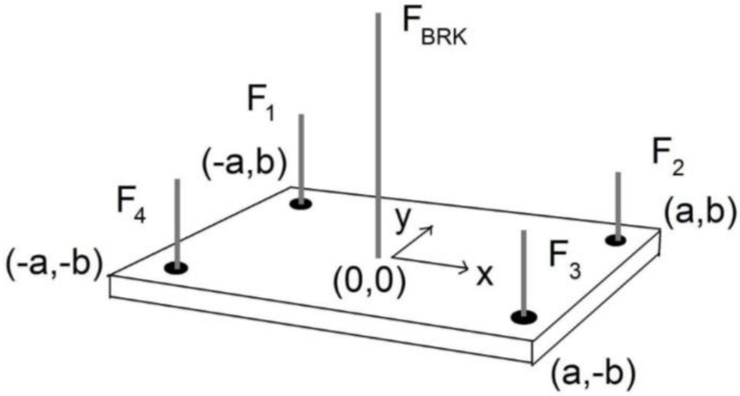


Figure 10 - Schematic representation of the operation of a force plate.

As shown above in Figure 10, the four force transducers are located at the coordinates (± a, ± b). For example, the force F_1 of the first sensor acts vertically at the location (-a, b), represented by a vertical gray bar. The zero point (0,0) is in the middle of the force plate. The force F_{BRK} represents the resulting floor reaction force vector. Its height is given by the sum of F_1, F_2, F_3 and F_4 (Schubert & Banzer, 2014).

Looking now at a living mass of a standing person, not a rigid mass, there are some non-trivial consequences. As correctly stated by Murray et al. (1967), the force transducers measure not only the weight of the mass, component-wise divided among the four sensors, but also the vertical components of those forces necessary to control the body center of mass (CoM) fluctuations. This implies that measurement using a force plate does not detect center of gravity (CoG) but a parameter called "center of pressure" (CoP) (Winter, 1995; Winter, Patla, & Frank, 1990).

The CoP can thus be defined as the point at which the ground reaction force vector, as the sum of all forces between the human body and the support surface, begins. For dynamic reasons the CoP must always overestimate the CoG and in addition to the larger amplitude also has an increased share in higher frequency ranges (Winter, 1995; Winter et al., 1990).

2.4 Sensorimotor training

In recent years, sensorimotor training or proprioceptive training has been given high priority in prevention and rehabilitation. Taube, Gruber, & Gollhofer (2008) describe this training as a training of postural control. However, there is no consensus on naming this type of training. The terminologies such as "balance training", "neuromuscular training", "proprioceptive training" or "sensorimotor training" are often used in the literature for the training of postural control. According to Gisler-Hofmann (2008), this training does not improve or change proprioception (the perception of joint positions or their changes, as well as the perception of movements), but rather a change in the spinal, supraspinal and cortical centers of the nervous system takes place. In these centers, afferent impulses are perceived and as a result motor correction and functional patterns can be planned (Gisler-Hofmann, 2008). Since motor performance is closely related to sensory information, this complex is summarized under the term sensomotoric. Roughly speaking, the field of sensorimotor activity consists of three major levels. The level of information, consisting of extrasensory (sensory perception) and proprioception. A processing level consisting of the spinal, supraspinal and cortical parts of the nervous system. The third level represents the execution level, which includes the motor system and the musculature. Accordingly, the term "sensorimotor training" is the most appropriate term.

Sensorimotor training promotes the regeneration of nerve cells, improves the sensitivity of the muscle spindles, supports the formation of dendrites, promotes neurobiological mechanisms and neuromuscular plasticity, improves sensorimotor learning (Gisler-Hofmann, 2008) and, thus, influences the same sensomotoric levels. Therefore, Gisler-Hofmann (2008) concludes that the neural adjustments resulting from the training can be used to carry out sensorimotor training in order to be able to actively stabilize joints, to improve global, local and also intramuscular coordination and thus to optimize strength. Since a human movement, however routine it may be, is never executed exactly in the same way. The central nervous system in the background must always include some deviation and uncertainty. The faster it is possible to react to such a variation, the smoother and safer the movements are. Fields of application for sensomotoric training are therefore firstly competitive, but also everyday situations in the form of balance improvement, fall prevention and injury prevention (Gisler-Hofmann, 2008). Taube et al. (2008) add that in addition an improvement in posture and jumping power can be achieved. They also point out that sensorimotor training for many different target groups such as athletes, children and the elderly as well as disabled persons can be applied.

2.5 Unstable shoe constructions

Shoes are typically designed to provide the user with stability and support while moving (Landry et al., 2010). Stability can be achieved either if the necessary stability is ensured by the shoe or is provided by activation of the stabilizing musculature itself. In order to achieve muscular stabilization certain exercises are often used, especially sensorimotor training. The exercises are usually performed on so-called balance boards, on soft mats or on unstable surfaces, such as a wobble board (Turbanski, Lohrer, Nauck, & Schmidtbleicher, 2011). In recent years, several special shoes have been developed that promise the same neuromuscular effects as sensorimotor training due to their unstable sole (Landry et al., 2010; Nigg et al., 2006; Romkes, 2008; Turbanski et al., 2011)

Nigg (2009), Landry et al. (2010) as well as Taniguchi, Tateuchi, Takeoka, & Ichihashi (2012) claim that these shoe designs, such as the MBT shoe (Masai Barefoot Technologie, shown in Figure 11 below), activate the small muscle groups around the ankle, which are neglected when wearing flat shoes. The activation and strengthening of these muscles, according to (Nigg, 2009) is especially desirable because these muscles are used for movement and movement control. With fast direction changes these small muscles provide stability in the ankle, which can play a significant role in the prevention of injury. In addition, manufacturers claim that this shoes can be used in a variety of ailments. Due to the special sole, the developers of these shoes claim that an improvement of back problems might be achieved by correcting the posture of the wearer. In addition, foot pathologies such as hallux valgus, flat foot, heel spurs and achilles tendon inflammations but also circulatory disorders can be positively influenced (Romkes, Rudmann, & Brunner, 2006). Most of these shoe designs are designed to give the wearer a barefoot feel, so the benefits of walking barefoot can also be harnessed while wearing shoes (Nigg, 2009). Therefore, especially the aspects of the foot shape, the special kinematics or the feeling of barefoot walking are taken up by the manufacturers, so (Nigg, 2009). These shoes are therefore often referred to as "barefoot shoes". However, this terminology is a bit misleading, since a state like walking barefoot cannot be given. Despite everything, there seems to be some benefit for the individual wearer (Nigg, 2009). For example, several studies examined the pressure distribution on the foot in standing and gait, the static and the dynamic balance as well as changes in kinetics and kinematics while walking with unstable sole constructions. Papalia et al. (2015) pointed to a change in these parameters that may be beneficial for improving posture and proprioception.

Figure 11 - MBT shoe with its typical rounded sole construction

The findings regarding muscle activity while walking are not consistent. Landry et al. (2010) found an increased activation of the flexor digitorum longus muscle and the anterior compartment musculature when initially wearing an unstable sole construction (MBT Shoe) compared to barefoot and a control shoe. In a second measurement after six weeks, the increased muscle activity of the above mentioned muscles was maintained and also differences in the muscle activity of the two mm. peronei could be detected while wearing the MBT shoe. However, the tests were only performed standing, so that a conclusion on the muscle activity during walking is not possible. Furthermore, the subjects were informed in advance about the handling of the shoe and trained, what they have to pay attention to while wearing the unstable shoe construction. Therefore, this result cannot be applied to everyday life, as for a similar result, an explanation would always be necessary.

Forghany, Nester, Richards, Hatton, & Liu (2014) also noted a difference in muscle activity between two similar unstable sole structures (both roll-over shoes) and control shoes during walking. In a single measurement, an average increased muscle activity of the m. soleus was found. Tendencies of increased activity could also be identified for the m. gastrocnemius medialis and the m. quadriceps rectus femoris. Furthermore, a reduced average as well as maximum muscle activity of the m. tibialis anterior was found.

Romkes et al. (2006) came up with comparable results regarding these muscles when comparing the MBT shoe to a reference shoe after four weeks of wearing. The subjects of this study were also instructed in advance regarding the walking technique with the MBT shoe. Forghany et al. (2014) and Romkes et al. (2006) only examined the difference between unstable sole construction and a control shoe. However, as these sole constructions aim to simulate a barefoot situation, further comparison with barefoot walking would be desirable.

Controversial to this is the study by Branthwaite, Chockalingam, Pandyan, & Khatri (2013). They did not notice a short-term effect on an unstable sole construction (MBT shoe) compared to a sports shoe. Rather, they describe very individual EMG results of the subjects. In contrast to comparable studies the participants were not familiarized with the handling of the MBT shoe in advance. The sports shoes used in the study as a reference shoe were brought by the study participants themselves, so the results of the comparison between the sports shoe and the unstable sole construction can only partially be considered as generally valid. Branthwaite et al. (2013) recommend individual assessments to determine the suitability of the shoe and its individual results. For people who react with an improvement in muscle activity the shoes could be used as a training device.

Granacher, Roth, Muehlbauer, Laser, & Steinbrueck (2011) also looked at an unstable sole construction (Biodyn sandal, shown in Figure 12) for possible effects on muscle activity in standing and gait. They found an increase in muscle activity of the m. peroneus longus as well as a higher activity of the m. gastrocnemius medialis compared to a control shoe in standing. When walking with the Biodyn sandal, the m. peroneus longus and the m. soleus tended to be more active than when walking with a reference shoe. However, there were no statistically significant differences in muscle activity between barefoot walking in the test shoe. Nevertheless, Granacher et al. (2011) suggested that the shoes could demand the balance and the associated muscle activity in everyday life and could contribute to the strengthening of the muscles.

Figure 12 - Biodyn sandals (Granacher et al., 2011)

Horsak & Baca (2013) also compared shoes with an unstable sole construction (Reebok Easy Tone, shown in Figure 13) with a reference shoe provided by the test subjects themselves. Muscle activity was recorded while walking, with no significant differences between the test shoe and the reference shoe. However, a slight increase in muscle activity of the vastus medialis and vastus lateralis of the m. quadriceps femoris was measurable. Horsak & Baca (2013) point out that increased muscle activity does not indicate clinical relevance. The test subjects were given the shoes as early as two weeks before the test in order to familiarize themselves with the unstable sole construction. While it has been suggested to the participants to wear the shoe as often as possible, no standardized number of hours has been established, so the results can only be considered partially valid.

Figure 13 - Reebok Easy Tone Reenew shoe model - cross-section profile (Horsak & Baca, 2013)

Price et al. (2013) compared the short-term effects of several unstable sole constructions to each other and to a reference shoe. All these shoes had increased muscle activity of the mm. peronei in common, an increased activity with respect to all other muscles near the intervertebral joints was very different from shoe to shoe. This is particularly evident since the muscle activities of this study were also assigned to the different gait phases and thus also the exact time of muscle activity could be assigned. Price et al. (2013) concluded that wearing unstable shoe constructions has an impact on muscle activity, but these effects depend on the sole shape and are very product specific.

2.6 X10D shoe construction

The X10D (pronounced "extend"), a new unstable sole construction, has been on the market since summer 2013. Through its novel concept the shoe should encourage the wearer to relocate the pressure when walking, especially on the lateral edge of the foot, as it is the case with barefoot walking. This should make human gait more economical, according to manufacturer and co-developer Swager van Dok. This is to be achieved by the construction of the sole. The sole consists of four different elements. The first element, the outsole, describes by its shape the part of the foot, which is optimally loaded during the rolling motion when walking barefoot. The second and fourth elements are harder and are therefore intended to distribute the pressure better. In addition, the position of the fourth element should provide as a proprioceptive input. The third element is the midsole, which lacks the medial part of the sole, causing the shoe to become unstable in the medio-lateral direction (see Figure 14). This instability then must be compensated by the wearer (Swager van Dok et al., 2015). According to the manufacturer, especially people with a lowered arch of the foot should profit from this concept. Swager van Dok et al. (2015) examined the pressure distribution with a pressure pad before and after eight weeks of wearing the X10D under three different shoe conditions - the X10D, a reference shoe and a barefoot situation. The results showed a change in pressure during walking with the X10D, which was similar to the pressure characteristics of barefoot walking, especially at the midfoot area. Swager van Dok et al. (2015) point out that the X10D could be an alternative method of treating foot and postural disorders, as these are often associated with insufficiently functioning foot muscles.

Figure 14 - The four sole elements of the X10D. 1. outsole, 2. harder guide element, 3. midsole with recesses on the medial side, 4. proprioceptive guide element (Swager van Dok, 2019)

3 Research questions and hypotheses

There are already various studies on the positive effects of several unstable sole constructions. However, the design of the X10D shoe differs from the previous sole constructions by the absence of the medial sole edge. Thus, this property is the indication for further investigation with this shoe, which has just been released on the market. Swager van Dak et al. (2015) already noted an improvement in pressure distribution during walking with the X10D. Another study of the University of Osnabrück, Germany dealt with the effects of the X10D on medial knee pain. A change in muscle activity while walking with these shoes was studied by two undergraduate students of the University of St. Pölten, Austria. However, a full investigation of the impact of these shoes with the specific unstable sole construction on the muscular activity of a large number of lower extremity muscles involved in the gait and the effects of this shoe on CoP and thus on the postural sway has not yet been carried out. This results in the following questions and hypotheses for this thesis:

1. *How does wearing the X10D shoe affect the maximum and average muscle activity of lower extremity muscles during gait in healthy adults aged between 18 and 65?*
2. *How does wearing the X10D shoe affect the postural sway when standing upright in healthy adults aged between 18 and 65?*
3. *How does wearing the X10D shoe affect postural sway when standing upright on one leg in healthy adults aged between 18 and 65?*

Referring to the first research question, the following hypotheses arise:

- *The first hypothesis is that there is a significant difference in maximum muscle activity of lower extremity muscles when wearing the X10D shoe, compared to a reference shoe and barefoot.*
- *The second hypothesis is that there is a significant difference in average muscle activity of lower extremity muscles when wearing the X10D shoe, compared to a reference shoe and barefoot.*

Referring to the second research question, the following hypothesis arise:

- *Wearing the X10D shoe effects the fluctuations of postural control significantly when standing upright compared to a reference shoe and barefoot.*
- *Wearing the X10D shoe effects the fluctuations of postural control significantly when standing upright on one leg compared to a reference shoe and barefoot.*

4 Methodology

To ensure the reproducibility of the used setup and for understanding the emerged challenges, the used requirements and the detailed measuring procedure is described in this chapter

4.1 Subjects

This study was carried out at one measuring session. Before the measurements were performed all of the 33 participants were informed about the benefits and intervention of this study and gave their written consent for voluntary participation. In Table 2 the baseline characteristics are shown.

Table 2 - Baseline characteristics

	N	Means ± SD	p-value
Male	14	-	-
Female	19	-	-
Age [yrs]	33	32,4 ± 12,7	0,000
Size [cm]	33	174,3 ± 8,8	0,248
Weight [kg]	33	70,7 ± 11,9	0,888
BMI [kg/m^2]	33	23,1 ± 2,6	0,614
Shoe size [EUR]	33	40,7 ± 2,6	0,268
Leg axis	33	-0,5 ± 1,1	0,000
Arch of the foot	33	-1,1 ± 1,4	0,001

The entire data collected was completely anonymised by a consecutive ID number. None of the selected participants had acute pain or an injury or surgery within the last six months on the lower extremities. Furthermore, the subjects were excluded from this study if they already had experience in wearing the X10D shoe. An ethical approval was obtained by the local ethics committee and all the subjects gave their written consent for voluntary participation.

4.2 Measuring equipment

The gait parameters were recorded with a pressure measuring plate and a surface EMG system. The wireless surface EMG system (Noraxon, United States of America, Scottsdale, Arizona; Research DTS) in combination with the adhesive ECG electrodes (Ambu, Denmark, Kopenhagen; BlueSensor SP) operated at 1500 Hz to record eight lower extremities muscles. Adhesive rings were used to stick the reference electrodes to the skin. The gait parameters such as pressure distribution, gait cycles, left/right recognition and so on were recorded by two pressure measuring plates (Zebris, Germany, Isny; FDM 1.5) embedded in a walkway, with a total length of three meters. The whole data output from the EMG and the pressure measuring plate were handled with the "Noraxon Myo Research XP Master Edition 3.9" software. To guarantee the same speed in every round measured two light sensitive barriers were put on the track, one at the beginning and one at the end of the pressure measuring plate. To prevent participants from cutting off the track and then subsequently putting on their first step in a different angle on the plate, a small pin was set up, to ensure that all participants followed the given route (Figure 15).

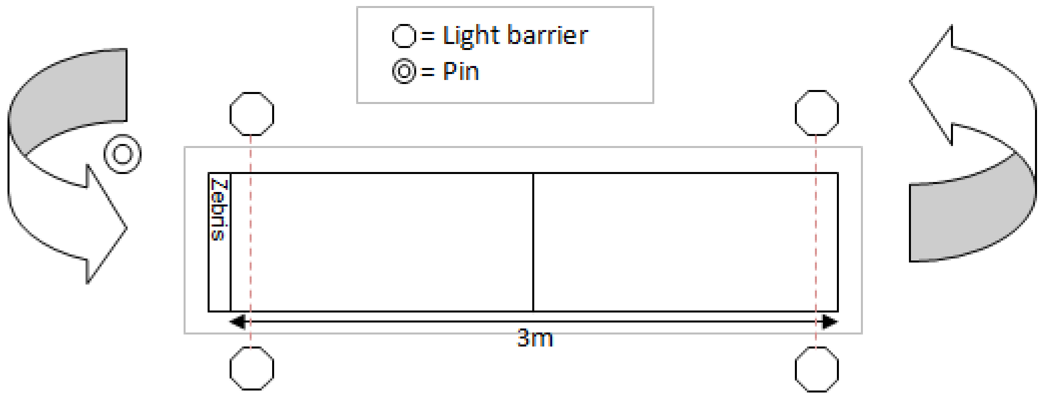


Figure 15 – A schematic depiction of the setup

The CoP parameters were recorded at 1000 Hz with a force plate (Kistler, Switzerland, Winterthur; type 9286BA). Furthermore, the Kistler charge amplifier, type 5691A, was necessary to convert the charge emitted by the piezoelectric sensors into a proportional voltage. The prior served as an input for analyse systems and further signal processing. This powerful data acquisition system (DAQ) was used with the Kistler data acquisition software BioWare Version 5.3.0.7, DataServer API (DataServer.dll), type 2873A. For standardizing the double leg stance of the participants, a handmade wooden block (Figure 16) was used, to ensure a constant distance between their feet. For standardizing the single leg stance, a piece of tape with some marker points on it was attached to the middle of the force plate.

Figure 16 – A wooden block was used to standardize the stride width.

4.3 Positioning of the surface EMG

Before the sensors and the electrodes were placed on the participants, their skin was prepared. First of all, the participants were asked to free the testing zone from clothing and to heighten their shirts if necessary. This was only the case if the shirt was too long and would come in contact with the cables of the sensors. The described scenario was performed to avoid disturbances as every irritation from outside could lead to artefacts in the records. For that reason, the participants were also asked to wear tight sports underwear only.

An proper skin preparation, shown in Figure 17, is a requirement to get a good quality of EMG signals. The two main goals of the preparation are a stable electrode contact and a low skin impedance. To get a stable electrode contact to the skin it is essential to remove the hair from the leg. In this study this was carried out by shaving an area on the measured leg with a disposable razor. After that a special abrasive and conductive cleaning paste (Everi, abrasive paste) was applied to remove dead skin cells and to generate a higher blood flow through rubbing the skin. To finalize the recommended procedure for dynamic measurements the treated skin area was cleaned with a skin disinfectant (Konrad, 2005).

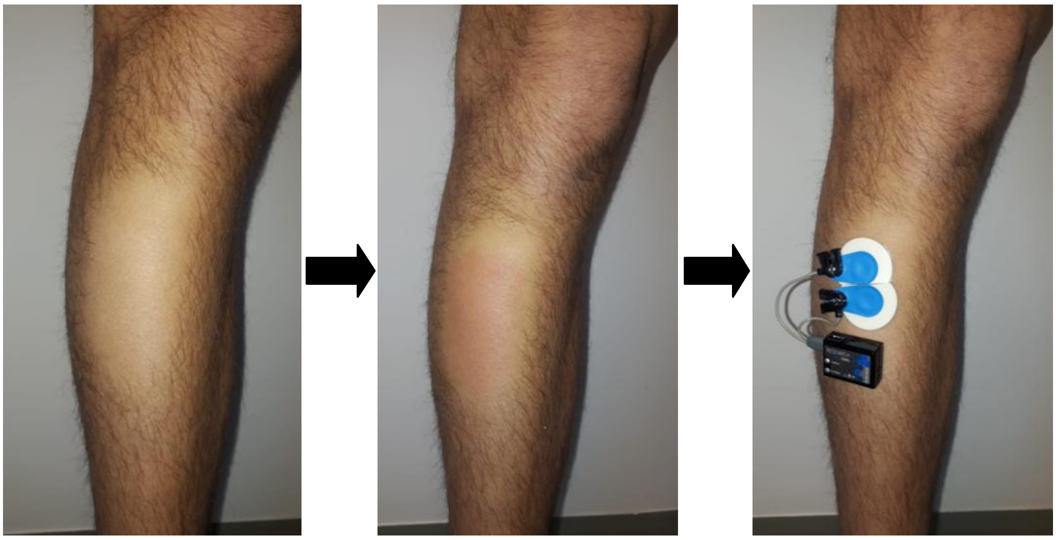


Figure 17 – A proper skin preparation, which ensures a stable electrode contact and a low skin impedance.

The second requirement to ensure reliable EMG signals is the positioning of the adhesive electrodes and the reference sensor. First the Ambu BlueSensor SP had to be cut to maintain an inter-electrode distance of 20 mm, right beside the centre of the sensor. Then the two edited sides were put together to obtain the recommended inter-electrode distance, shown in Figure 18.

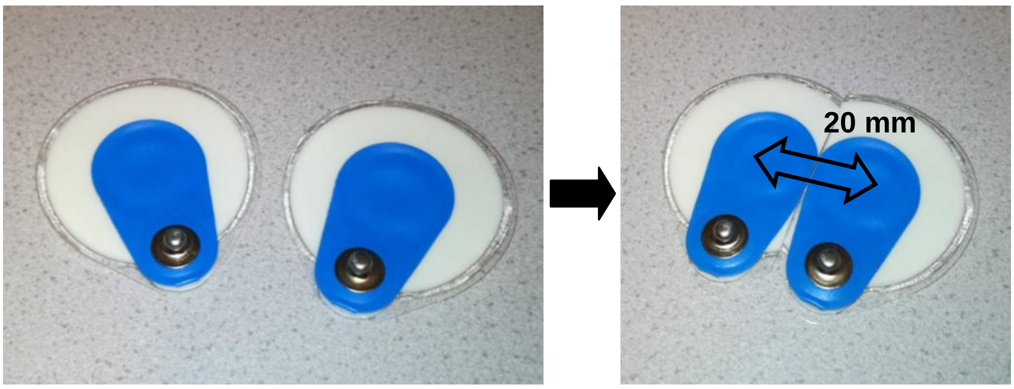


Figure 18 - In order to obtain the two cm inter-electrode distance it is necessary to cut the electrodes.

In this study the palpation and the applying of the electrodes is peformed by a single rater. As recommended in literature, such measurements should be performed by a single rater, rather than by different raters, in terms of reliability (Moriguchi et al., 2009). To improve the interrater reliability, the electrodes were attached according to the SENIAM guidelines. The eight electrode pairs were attached to eight different muscles (m. gluteues maximus, m. gluteus medius, m.

biceps femoris, m. gastrocnemius medialis, m. peroneus longus, m. tibialis anterior, m. quadriceps femoris vastus medialis, m. tensor fasciae latae).

1. For attaching the electrodes on the **m. gluteus maximus** the participants were asked to lie down in a prone position on the practitioner table. The electrodes were placed at 50% on the line between the sacral vertebrae and the greater trochanter. This position corresponds with the greatest prominence of the middle of the buttocks well above the visible bulge of the greater trochanter. The orientation was set to be in the direction of the line from the posterior superior iliac spine to the middle of the posterior aspect of the thigh.
2. For attaching the electrodes on the **m. gluteus medius** the participants were asked to lie down on the side on the practitioner table. The electrodes were placed at 50% on the line from the crista iliaca to the trochanter. The orientation was set to be in the direction of the line from the crista iliaca to the trochanter.
3. For attaching the electrodes on the **m. biceps femoris** the participants were asked to lie down in a prone position on the practitioner table with their knee flexed to less than 90 degrees with a slight lateral rotation. The electrodes were again placed at 50% on the line between the ischial tuberosity and the lateral epicondyle of the tibia. The orientation was set to be in the direction of the line between the ischial tuberosity and the lateral epicondyle of the tibia.
4. For attaching the electrodes on the **m. gastrocnemius medialis** the participants were asked to lie down in a prone position with their knee extended and the foot projecting over the end of the practitioner table. The electrodes were placed on the most prominent bulge of the muscle. The orientation was set to be in the direction of the leg.
5. For attaching the electrodes on the **m. peroneus longus** the participants were asked to sit down with their thigh medially rotated. The electrodes were placed at 25% on the line between the tip of the head of the fibula to the tip of the lateral malleolus. The orientation was set to be in the direction of the line between the tip of the head of the fibula to the tip of the lateral malleolus.
6. For attaching the electrodes on the **m. tibialis anterior** the participants were asked to sit down on the practitioner table. The electrodes were placed at a third on the line between the tip of the fibula and the tip of the medial malleolus. The orientation was set to be in the direction of the line between the tip of the fibula and the tip of the medial malleolus.

7. For attaching the electrodes on the **m. quadriceps femoris vastus medialis** the participants were asked to sit down on the practitioner table with their knees in slight flexion and the upper body slightly bend backwards. The electrodes were placed at 80% on the line between the anterior spina iliaca superior and the joint space in front of the anterior border of the medial ligament. The orientation was set to be almost perpendicular to the line between the anterior spina iliaca superior and the joint space in front of the anterior border of the medial ligament.
8. For attaching the electrodes on the **m. tensor fasciae latae** the participants were asked to lie down on the side on the practitioner table. The electrodes were placed on the line from the anterior spina iliaca superior to the lateral femoral condyle in the proximal sixth. The orientation was set to be in the direction of the line from the anterior spina iliaca superior to the lateral femoral condyle.

All the electrode pairs had their own reference electrode which was placed on the backside of the transmitter. This transmitter was placed in a position that the cables would touch each other as little as possible. If the cables would touch each other too much, signals could be irritated and this could lead to artefacts in the records.

4.4 Study procedure

After welcoming the participants, they were asked to fill out an information sheet concerning the baseline characteristics (sex, age, size, weight, shoe size, dominant leg), sign the consent form and listen carefully to the initial explanation given. Certain points on the information sheet, such as leg axis and feet arch of the participants, had to be filled out by the examiners. These two points have been rated by visual inspection of two physical therapists and were categorized on a scale from -5 to +5. Whereas -5 stands for valgus in the knee joint respectively pronation in the ankle joint and +5 for varus in the knee joint respectively supination in the ankle joint. Afterwards the weight of the norm shoe and the participants average walking speed were also registered by the examiners and noted on the information sheet. Subsequently, the participants were asked to change their clothes, as explained above. All 33 participants indicated their right leg as the dominant leg, on which the measurements were performed.

As described in section 4.3, an important first step was the proper preparation of the skin. Primarily the dominant leg of the participants has been shaved in order to avoid an instable electrode contact. After the shaving, the dead skin cells of the superficial layer of the skin were removed by applying a special abrasive and conductive paste. To eventually clean the skin from the paste, a skin disinfectant was used.

After completing the skin preparation, the adhesive electrodes and the reference electrode were attached to the skin and connected with the appropriate cables. To keep the reliability of palpation and attaching the electrodes as high as possible the whole hands-on procedure was carried out by a single examiner. The electrodes were attached according to the SENIAM guidelines, described in detail in section 4.3.

After a successful positioning of the eight electrode pairs, the EMG signal was checked in the software. If a good quality of the EMG signal was given, the examiners started to explain the maximum voluntary contraction (MVC) concept and the associated measuring to the participants. Each muscle function test for each of the eight muscles was explained in detail. The order and the execution of each muscle function test are explained at the end of this paragraph. The participants were asked to generate a muscle tension as high as possible against the resistance given. As soon as the procedure was clear to the participant, the muscle function tests according to Hislop & Montgomery, (2007) were started:

1. For testing the **m. gluteus maximus** the participants were asked to lie down in a prone position on the practitioner table with their knees flexed in a 90-degree angle. The rater’s hand, causing the resistance, was placed proximal of the knee. The other hand was placed on the pelvis to restrict the continuous movements. The participants were asked to raise the foot up to the ceiling and to keep the knee in the flexed position.
2. For testing the **m. gluteus medius** the participants were asked to lie down on their side on the practitioner table with their tested leg upwards. The rater’s hand, causing the resistance, was placed lateral on the knee. The participants were asked to raise the leg straight up and not to give in to the resistance.
3. For testing the **m. biceps femoris** the participants were asked to lie down on the practitioner table in a prone position with the tested knee flexed in less than a 90-degree angle. The leg was slightly rotated outwards and the toes pointed laterally. The rater’s hand, causing the resistance, was placed on the inside of the ankle. The participants were asked to flex the knee against the resistance and not to give in to the resistance. The toes still pointed laterally.

4. For testing the **m. gastrocnemius medialis** the participants were asked to stand only on the tested leg with their knee extended. They were allowed to hold on to the practitioner table with two fingers. Then the participants should push up on their tiptoes and get back down. Those calf raises were carried out a few times.
5. For testing the **m. peroneus longus** the participants were asked to sit down on the practitioner table with their knees in flexion and their upper body slightly bend backwards. The tested foot was placed on the rater's femur. The rater's hand, causing the resistance, was placed on the dorsal and lateral part of the forefoot. The participants were asked to rotate the foot downwards and outwards. They should keep the muscle contraction high and not give in to the resistance.
6. For testing the **m. tibialis anterior** the participants were asked to sit down on the practitioner table with their knees in flexion and the upper body slightly bend backwards. The tested foot is placed on the rater's femur. The rater's hand, causing the resistance, was placed on the dorsal and medial part of the foot. The participants were asked to rise the foot upwards and inwards. They had to keep the muscle contraction high and not give in to the resistance.
7. For testing the **m. quadriceps femoris vastus medialis** the participants were asked to sit down on the practitioner table with their knees in flexion and the upper body slightly bend backwards. The rater's hand, causing the resistance, was placed on the ankle. The participants were asked to extend the leg and not to give in to the resistance.
8. For testing the **m. tensor fasciae latae** the participants were asked to lie down on the practitioner table on their side with their tested leg upwards with their hip flexed in a 45-degree angle. The rater's hand, causing the resistance, was placed lateral on the knee. The participants were asked to raise the leg straight up and to not give in to the resistance.

For all the muscles tested, the participants were cheered on to contract the muscle as much as possible.

Following to the execution of the MVC measuring the examiners started to explain the further process. The next step was to choose the order of the measuring situations as well as the shoe situations. To guarantee a valid randomization this task was carried out with the support of random.org. The two measurement situations, gait and stance, were assigned to numbers and were drawn first. This was followed by the draw of the three shoe situations (barefoot, norm shoe, X10D shoe). Both measurement situations were carried out with the shoe, drawn first. Afterwards the participants carried out the same procedure with the two remaining shoe situations. So, the participants did not have to change their shoes unnecessarily often.

Starting with the gait measuring the participants had up to three minutes to get familiar to the track and especially to find their self-paced velocity. They were asked to walk as usual as possible and walk at their preferred speed. During the familiarization phase the walking speed and the quality of the EMG signals were monitored (Figure 19).

Figure 19 – An example of a good quality EMG signal without artefacts and outliers

If the chosen speed did not vary over the period of three trials, the recording was started without telling the participants. The variation in speed was chosen to a limit of 10% upwards and 10% downwards of the determined average individual walking speed. By not telling the participants the starting point, normal gait patterns as possible in a laboratory situation were expected. During the recording, the examiners checked whether the participants were stepping properly on the pressure measuring plate and whether the EMG signals looked fairly similar to the standard muscle activity pattern. The participants had to walk until the recording was stopped. After completing five valid trials the recording was stopped and the participants continued with the stance analysis.

In standing they had to complete ten trials with a duration of 20 seconds each. The first five trials were performed with double leg support and the second five with single leg support. As with the gait analysis, only the dominant leg was measured for the single leg stance. To standardize the double leg stand, a wooden block was used which was placed between the feet of the participants. This should avoid a variation of their step width and keep their feet in the same position. The block was only used for the standardization and was removed before the measurement started. After each completed trial the participants had to step off the plate and get then back on it with a new standardization procedure, so the required resetting of the plate could have been carried out. During the measuring the participants were asked to let their arms hang loose and look forward on a certain marker point placed on the wall in two meters in front of them and on the participants eye level. Furthermore they were not allowed to move for the 20 seconds of recording. In the double leg stand measuring, all of the participants could keep their balance so none of the trials had to be discarded and subsequently repeated.

During the single leg stance the standardization was given by an adhesive tape (Figure 20) which was placed in the middle of the plate. On this tape were some marker lines and a cross in the middle where the participants should put the center of their foot. They were asked to remember this position and repeat the standing position as good as possible for the next trials.

Figure 20 – Force plate with adhesive tape for one leg stand standardization

The arms should be again in a hanging loose position but during the single leg stance they were allowed to move them for keeping the balance. The lifted leg, in this study always the left leg, should be in a flexed position, which was determined by a 90-degree angle in the hip joint and a 90-degree angle in the knee joint (Figure 21). The participant's line of sight should be again straight to the marker point in front of them. The measuring only started if the participants were able to hold the starting position stable for three seconds. As mentioned above, slight movements were allowed, since they were not completely avoidable, but they should always try to keep the explained starting position as good as possible.

Figure 21 – Standardized starting position in the single leg stance

In the single leg stance some of the trials had to be discarded and subsequently repeated. In most of the discarded trials the participants were not able to keep the balance over the time of 20 seconds. A trial was immediately cancelled if the participants had to step down with their left foot. Due to some repetitions, the measuring in the single leg standing took a bit longer but all of the participants were able to complete five valid trials. The same procedure was then carried out with the two other shoe situations.

After completing the measurements, the subjects were asked about the properties of the shoes as well as their first impressions when walking and standing.

4.5 Data processing

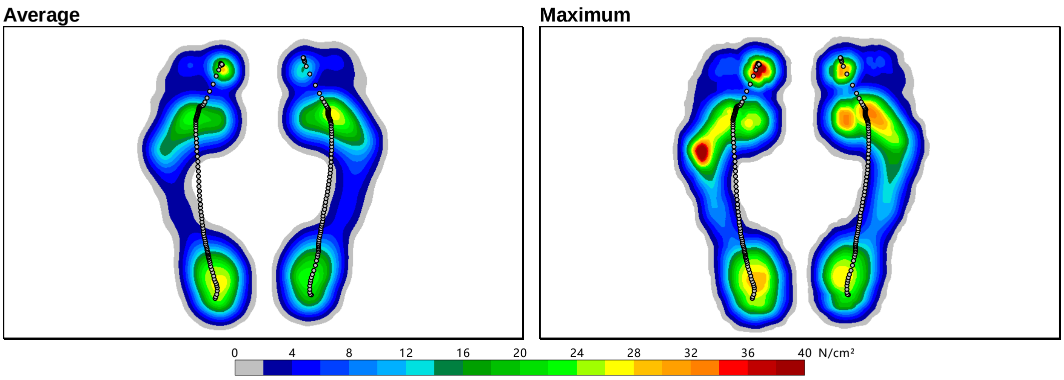
4.5.1 Gait data

The software "Noraxon Myo Research XP Master Edition 3.9" was used to record and process the data. The processing of the EMG raw signals included rectification, smoothing and amplitude normalization. Konrad (2005) describes the rectification the negative amplitudes are turned positive by being flipped up.

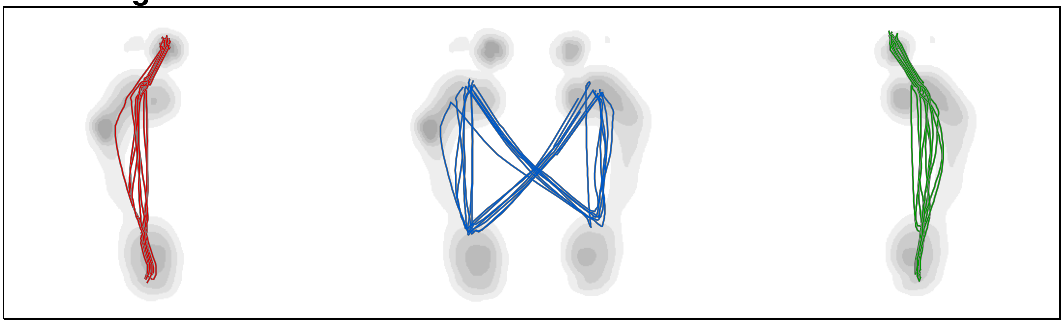
This makes it easier to detect the amplitudes and to calculate the standard amplitude parameters such as mean, minimum, maximum and integral. Digital smoothing minimizes the signal immanent variability by eliminating the non-reproducible amplitude peaks and as a result it is displaying the averaged signal trend of the amplitudes. This method is based on two algorithms. The Root Mean Square (RMS), which reflects the average signal of the amplitudes, is the currently recommended smoothing algorithm. The algorithm is determined for a specific time window, which typically ranges from 20 milliseconds (ms) for kinesiological studies in which fast movements such as jumps are recorded and goes up to 500 ms for slow or static motions. For most EMG experiments, a time window between 50 ms and 100 ms is sufficient because the larger the time window, the greater the risk of phase shifting in strongly increasing contractions gets. Due to the reason that walking is a slower activity, a time window of 80 ms was chosen for smoothing the recorded data. The third step in data preparation is the amplitude normalization, which aims to eliminate the influence of local divert conditions (Konrad, 2005). The most well-known amplitude normalization, also used in this study, is the already described MVC normalization. This method uses the maximum voluntary contraction of the individual muscles as a reference and indicates the measured muscle activity normalized during walking as a percentage of that value [% MVC].

Afterwards the raw data had to be checked manually for any artefacts in the records which were subsequently discarded. Not only the artefacts had to be eliminated but also incorrect first steps on the pressure measuring plate. Sometimes the participants also placed their right foot too far on the right side of the plate and stepped over the edge of the plate. These steps were removed as well. After preparing the raw data the especially designed reports in the Noraxon software were used. In this study the “MyoPressure Bilateral Gait Side Overlay Report” was used which is shown in Figure 22, Figure 23 and Figure 24. This report automatically compiles a left to right comparison of standard gait analysis parameters, time normalized averaged curves, CoP analysis and optional video analysis. Unlike the Bilateral Gait Report, it automatically creates a left-right curve overlay graph with a selection of amplitude parameters. Another advantage is that the steps are automatically detected, and no period definition settings are needed. Each stance phase and swing phase of each detected stride is marked in red (left side) and green (right side) bars. The report contains the pressure prints, the CoP gait line, a butterfly diagram, CoP parameters, a gait phase diagram, gait phase-, spatial- and time parameter table and the average force curves. The left- and the right-side curves were overlaid and analysed with a minimum, a maximum and a mean value. The MyoMotion averaged EMG curves are added automatically if measured in multi device setup configuration.

Pressure Prints



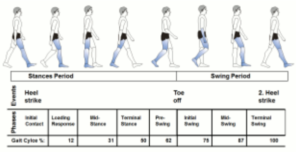
COP Diagram



COP Parameters

Länge der Ganglinie, mm	Links	209±2
	Rechts	210±4
	Diff, %	0.1
Single support Linie, mm	Links	127±6
	Rechts	109±8
	Diff, %	-13.8
Ant/Post Position, mm		117±19
Laterale Symmetrie, mm		-3±25

Figure 22 – On the first page of the MyoPressure Bilateral Gait Side Overlay Report the average Pressure Prints, the CoP Diagram and the averaged CoP Parameters of at least five valid trials are shown. The CoP Diagram and the CoP parameters are respectively shown for the right (green) and the left side (red).



Gait Phase Parameters

Standphase, %	Links	65.3±1.4
	Rechts	63.9±0.9
	Diff, %	-2.1
Gewichtsübernahme, %	Links	13.4±0.9
	Rechts	14.1±1.2
	Diff, %	5.3
Mittelstandphase, %	Links	36.7±1.1
	Rechts	36.1±1.4
	Diff, %	-1.6
Vorschwung, %	Links	15.2±1.7
	Rechts	13.7±1.0
	Diff, %	-9.7
Schwung Phase, %	Links	34.7±1.4
	Rechts	36.1±0.9
	Diff, %	3.9
Zweibeinstand, %		28.1±2.0

Gait Spatial Parameters

Fußrotation, Grad	Links	9.7±1.2
	Rechts	11.6±1.2
Schrittlänge, cm	Links	61±1
	Rechts	61±1
	Diff, %	-0.9
Länge Doppelschritt, cm		122±2
Schrittbreite, cm		5±2
Geschwindigkeit, km/h		4.0±0.1

Gait Time Parameters

Schrittzeit, sec	Links	0.55±0.01
	Rechts	0.54±0.01
	Diff, %	-0.3
Zeit Doppelschritt, sec		1.09±0.01
Kadenz, Schritte/min		110±1

Figure 23 – On the second Page of the MyoPressure Bilateral Gait Side Overlay Report the Gait Phase Paramaters, the Gait Spatial Parameters and the Gait Time Parameters are respectively shown for the right (green) and the left side (red). All of the values are the mean of at least five valid trials.

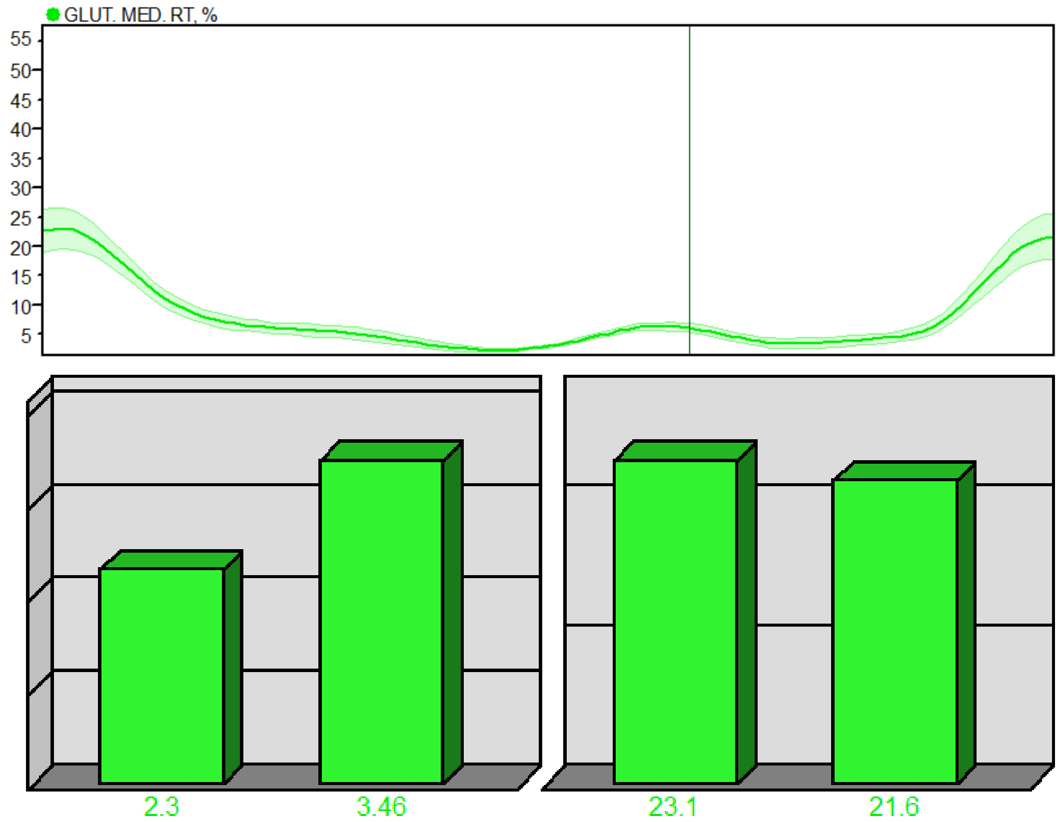


Figure 24 – An example of a random average EMG curve on the third page of the MyoPressure Bilateral Gait Side Overlay Report. On top of the graph the mean muscle activation in percentage of MVC (green line) with standard deviation (light green shade) of the m. gluteus medius for one gait cycle is shown. On the bottom of the graph the two bars on the left side show the respective average muscle activity in percentage of MVC of the entire stance and swing phase. The two bars on the right side show the associated peak values in percentage of MVC, also respective for the stance and swing phase.

To process the data, it was further necessary to export the data from the Noraxon software to Microsoft Excel by means of .slk files. For this study and consequently to be able to handle this huge amount of data it was necessary to import all the data into Matlab. Creating a code for representing the data was a big part of this study. Unfortunately .slk files do not operate with Matlab so first they had to be converted to .xlsx files manually. After importing the data and completing the data processing in Matlab. All of the data was exported back to an Excel file which contains the entire data recorded. Beside the two main variables, mean and peak activation of each muscle, eight time-distance variables were recorded:

- stance phase in %
- double support phase in %
- foot rotation in degree

- stride length in cm
- stride width in cm
- double stride length in cm
- velocity in km/h
- cadence in steps/min

These variables were also recorded for each shoe situation to see if there was any variation. For all variables the mean value of at least five valid trials was calculated.

The data analysis was performed with IBM SPSS Statistics, which was also used to remove certain outliers out of the records. By means of box plots those outliers were identified. Outliers may be erroneous data resulting from measurement errors, instrument failure or similar problems. A distinction is made between mild and extreme outliers. Mild outliers have an interquartile range (IQR) to the 1st or 3rd quartile from 1,5* IQR to 3,0* IQR and are represented as a circle in the box plot. Extreme outliers are more than 3,0 * IQR apart and are represented as a star in the box plot. Mild and extreme outliers should be and were excluded from further analysis.

4.5.2 Stance data

A force plate uses either strain gages or piezoelectric quartz crystals to convert force into electrical signals. Piezoelectric sensors form the basis of every Kistler force measuring system. With their advantage of high natural frequency and low susceptibility to interference, they provide the perfect characteristics for measuring dynamic, highly sensitive processes in biomechanics. Vertical, transversal and horizontal forces can be recorded with such a force plate, which is plotted in Figure 25.

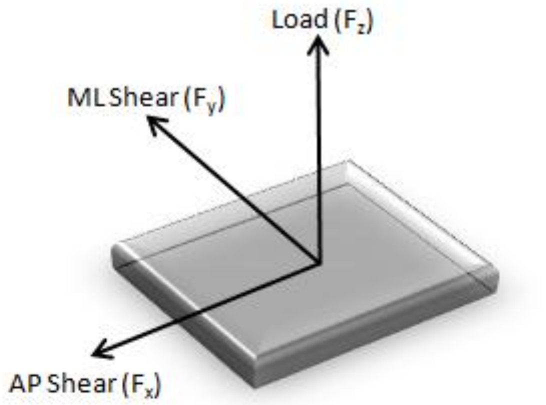


Figure 25 – According to Duarte & Freitas (2010) force plates measure three acting forces, the vertically acting force (F_z), the anterior-posterior shear forces (F_x) and the medial-lateral shear forces (F_y).

Due to recent literature we decided to record the CoP for double and single leg standing on the force plate. As mentioned in section 4.4 the participants had to carry out ten valid trials on the plate - five times double leg and five times single leg. The recording was carried out with a rate of 1.000 Hertz over a period of 20 seconds, this resulted in a sample length of 20.000. For double leg standing the ground reaction force is almost vertical, constant and corresponds to the opposite of body weight. An entirely different picture was shown in the plots of the single leg standing. Due to the lack of stabilization far more medio-lateral and anterior-posterior movements were detected. If an error occurred in the records, that certain trial was deleted and subsequently repeated. The recorded signals were exported from the BioWare software by means of .txt files. For completing the stance analysis, the data was analysed using a Matlab function based on the calculations described in Prieto, Myklebust, Hoffmann, Lovett & Myklebust (1996).

The following variables were calculated in the function and used for the evaluation:

- mean distance anterior-posterior (MDISTap)
- mean distance medio-lateral (MDISTml)
- root mean square distance anterior-posterior (RDISTap)
- root mean square distance medio-lateral (RDISTml)
- total excursion anterior-posterior (TOTEXap)
- total excursion medio-lateral (TOTEXml)
- mean velocity anterior-posterior (MVELOap)

- mean velocity medio-lateral (MVELOml)
- range anterior-posterior (RANGEap)
- range medio-lateral (RANGEml)

This function is based on the following equations (Prieto et al., 1996) to calculate the parameters:

The mean distance-AP ($MDIST_{AP}$) is the mean absolute value of the AP (anterior - posterior) time series and represents the average AP distance from the mean COP

$$MDIST_{AP} = 1/N \sum |AP[n]|$$

The root mean square distance-AP ($RDIST_{AP}$) from the mean COP is the standard deviation of the AP time series.

$$RDIST_{AP} = \left[{}^{1}/_{N} \sum AP[n]^2\right]^{1/2}$$

The total excursions-AP ($TOTEX_{AP}$) is the total length of the COP path in the AP direction, and is approximated by the sum of the distances between consecutive points in the AP time series.

$$TOTEX_{AP} = \sum_{n=1}^{N-1} |AP[n+1] - AP[n]|$$

The mean velocity-AP ($MVELO_{AP}$) is the average velocity of the CoP in the AP direction.

$$MVELO_{AP} = \ TOTEX_{AP}/T$$

The range is the maximum distance between any two points on the CoP path. The range-AP is the absolute value of the difference between the smallest and largest values in the AP time series (Prieto et al., 1996).

Every measure defined for the AP time series is similarly defined for the ML time series.

The results of the calculated parameters were stored in a Matlab struct for an easier processing. The entire data were separated by double and single leg stance. Thereafter, the collected stance data were merged with the gait data in an Excel (2016) spreadsheet version, to perform the statistical analysis more efficiently.

After the participants completed all the measurements, they were questioned about their subjective feelings while walking with the X10D and asked for a personal feedback on the shoes.

4.6 Statistics

The data analysis was performed with IBM SPSS Statistics Version 25. The alpha level was 5%.

In this study, the Shapiro Wilk test was set a priori to Kolmogorov Smirnov test due to its higher test strength. Apart from the variables of the parameter “age”, all other variables were normally distributed.

In this study the ANOVA (analysis of variances) with repeated measures was used to compare the mean values of the different shoe situations, although not all metric data were normally distributed, as mentioned above. Vasey & Thayer (1987) have shown that the ANOVA with repeated measures is relatively robust to violations of normal distribution assumption, especially if no further assumption has been violated (Berkovits, Hancock, & Nevitt, 2000). In addition, the sample size was higher than 25 ($n = 33$), which can additionally tolerate this assumption violation. The sphericity, which tests the homogeneity of the variances, was checked by using the Mauchly test. For a not assumed sphericity ($p = <0.05$) the Greenhouse-Geisser correction was used.

In case of significant differences, post hoc tests were used to identify which variables differ from each other. The post hoc tests provide paired averages comparisons of which averages differ significantly from each other. To counteract the alpha error cumulation the LSD (least significance difference) correction was used. Since it is an exploratory study, we decided to choose this minor correction and deliberately risked an alpha error, which means the null hypothesis gets rejected and a difference is accepted even though none exists.

5 Evaluation Results

The results of the statistical evaluation are described and graphically presented in the following chapters.

5.1 Results of the gait analysis

5.1.1 Results of the peak values

For the comparison of the peak values an ANOVA with repeated measures was carried out. As shown in Figure 26, two of the eight muscles indicate a significant difference in EMG peak values. The differences for the averaged peaks of the three different shoe situations are significant for the m. biceps femoris (p = 0,047) and even highly significant (p = 0,002) for the m. tibialis anterior. The highest peak values were determined on the norm shoe for the m. biceps femoris and on the X10D shoe for the m. tibialis anterior. The results are shown in Table 3.

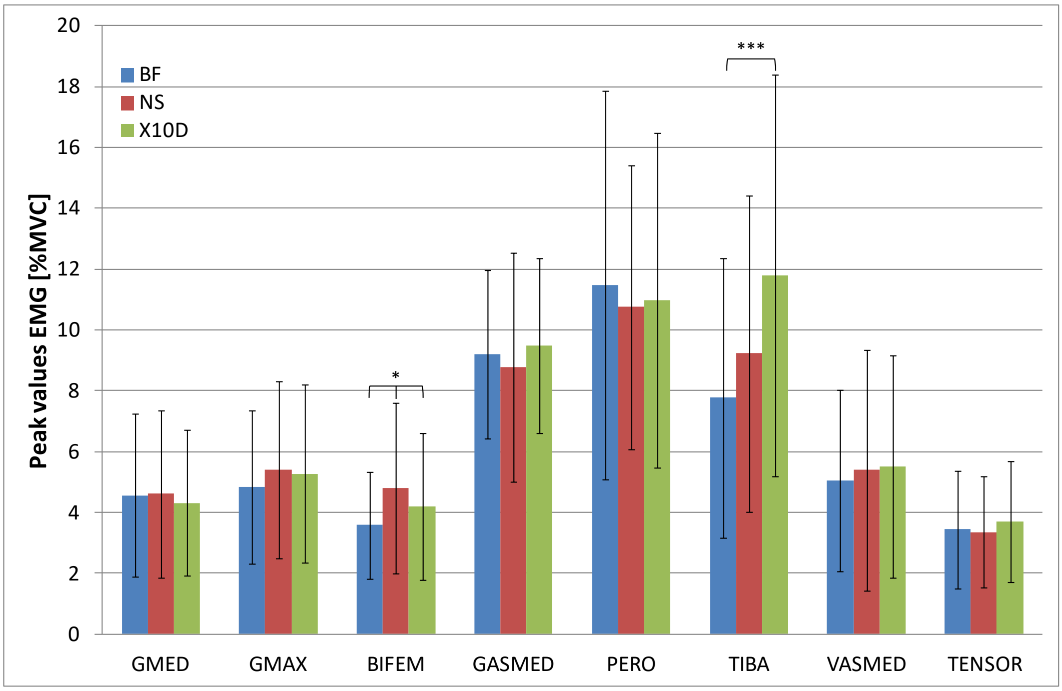


Figure 26 - The averaged peak values of the EMG signals of all participants with the associated standard deviation for each muscle and each shoe situation.

Table 3 – The averaged peak values of the EMG signals of all participants with the associated standard deviation for each muscle and each shoe situation.

EMG peak values									
Parameter	BF		NS		X10D			ANOVA	
[%MVC]	Mean	SD	Mean	SD	Mean	SD	df	F	p
GMED	4,41	2,62	4,52	2,78	4,31	2,43	2	0,200	0,819
GMAX	4,21	2,04	4,48	2,59	4,46	1,91	2	0,274	0,690
BIFEM	3,28	1,62	4,39	2,78	3,72	1,87	2	3,567	**0,047**
GASMED	8,86	2,59	8,53	3,71	9,56	2,95	2	2,059	0,137
PERO	10,25	5,37	10,49	4,81	10,91	5,52	2	0,440	0,613
TIBA	7,60	3,94	9,12	5,35	10,69	6,13	2	7,221	**0,002**
VASMED	5,04	3,03	5,39	4,03	5,51	3,72	2	0,581	0,533
TENSOR	3,31	1,92	3,21	1,84	3,35	1,83	2	0,224	0,748

Concerning the two significant p-values further statistics in form of pairwise comparisons (post hoc test) were carried out. In the case of the biceps femoris the significant difference could only be determined in the overall comparison and not in the pairwise comparisons, shown in Table 4. None of the three different shoe situations showed a significant difference to one of the other shoe situations. Apart from the pairwise comparisons, the partial eta squared (η^2) for the m. biceps fermoris amounts to a value of 0,121.

Table 4 - Pairwise comparisons of the different shoe situations concerning the peak activation of m. biceps femoris

Shoe situation		Mean difference [% MVC]	Standard error	Sig.	95% confidence interval for the difference	
					lower limit	upper limit
BF	**NS**	-1,111	0,515	0,112	-2,169	-0,053
	X10D	-0,440	0,340	0,265	-1,139	0,259
NS	**BF**	1,111	0,515	0,112	0,053	2,169
	X10D	0,671	0,382	0,236	-0,114	1,456
X10D	**BF**	0,440	0,340	0,265	-0,259	1,139
	NS	-0,671	0,382	0,236	-1,456	0,114

Different results are shown in Table 5, in which the pairwise comparisons for the m. tibialis anterior are described. The X10D shoe (X10D) shows a highly significant difference (p = 0,001) in comparison to barefoot (BF) and a tendency towards significance (p = 0,07) in comparison to the norm shoe (NS). The norm shoe in comparison to barefoot also indicates a tendency towards significance (p = 0,068). The highest average peaks in the EMG recordings were determined for the X10D shoe, followed by the norm shoe and barefoot. Apart from the pairwise comparisons, the partial eta squared (η^2) for the m. tibialis anterior amounts to a value of 0,211.

Table 5 - Pairwise comparisons of the different shoe situations concerning the peak activation of m. tibialis anterior

Shoe situation		Mean difference [% MVC]	Standard error	Sig.	95% confidence interval for the difference	
					lower limit	upper limit
BF	**NS**	-1,515	0,798	0,068	-3,152	0,122
	X10D	-3,085	0,805	0,001	-4,738	-1,433
NS	**BF**	1,515	0,798	0,068	-0,122	3,152
	X10D	-1,571	0,832	0,070	-3,278	0,137
X10D	**BF**	3,085	0,805	0,001	1,433	4,738
	NS	1,571	0,832	0,070	-0,137	3,278

5.1.2 Results of the mean values

To compare the mean values an ANOVA with repeated measures was calculated. None of the parameters in Table 6 show significant differences in the mean activation of each muscle. There were slightly differences between the three shoe situations which are shown in Figure 27 and also in Figure 28 over the entire gait cycle. No further statistics in form of pairwise comparisons were carried out.

Table 6 - The averaged mean values of the EMG signals of all participants with the associated standard deviation for each muscle and each shoe situation.

EMG mean values									
Parameter	BF		NS		X10D			ANOVA	
[%MVC]	Mean	SD	Mean	SD	Mean	SD	df	F	p
GMED	1,76	1,23	1,87	1,37	1,69	1,05	2	0,551	0,579
GMAX	1,69	1,46	1,58	1,31	1,66	1,16	2	0,383	0,684
BIFEM	1,13	0,58	1,33	0,68	1,22	0,68	2	1,196	0,310
GASMED	4,71	1,61	4,51	1,94	4,94	2,04	2	1,781	0,177
PERO	5,89	3,16	5,74	2,98	5,25	2,53	2	1,34	0,270
TIBA	3,23	1,94	3,50	1,88	3,86	2,24	2	1,790	0,177
VASMED	1,60	0,96	1,72	1,25	1,74	1,15	2	0,665	0,518
TENSOR	1,59	0,81	1,60	0,95	1,62	0,89	2	0,074	0,929

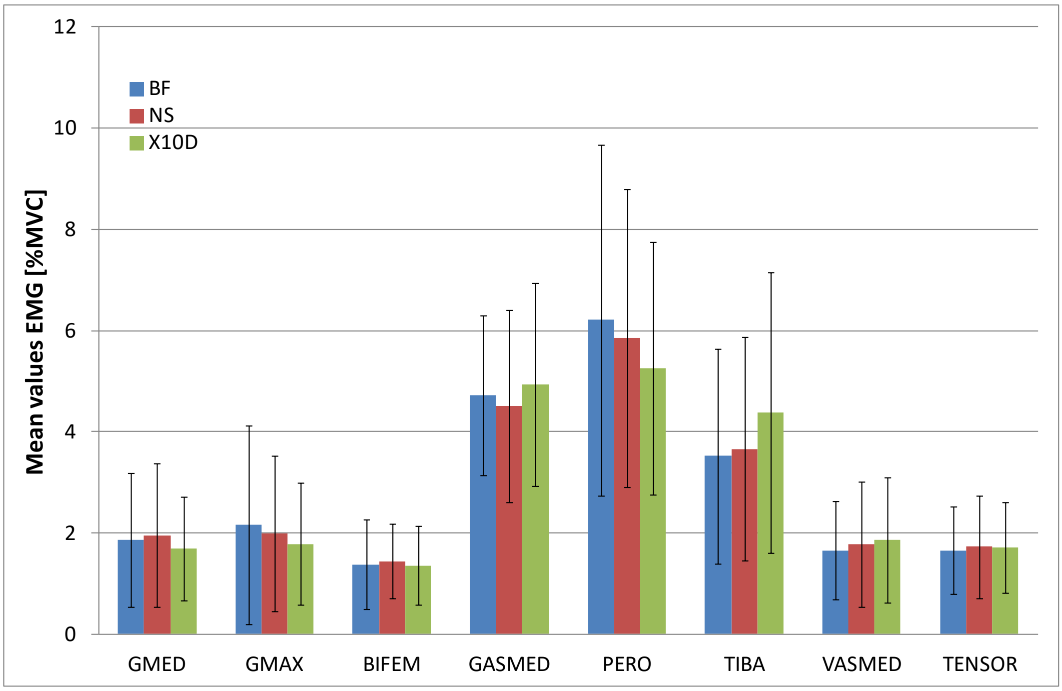


Figure 27 - The averaged mean values of the EMG signals of all participants with the associated standard deviation for each muscle and each shoe situation.

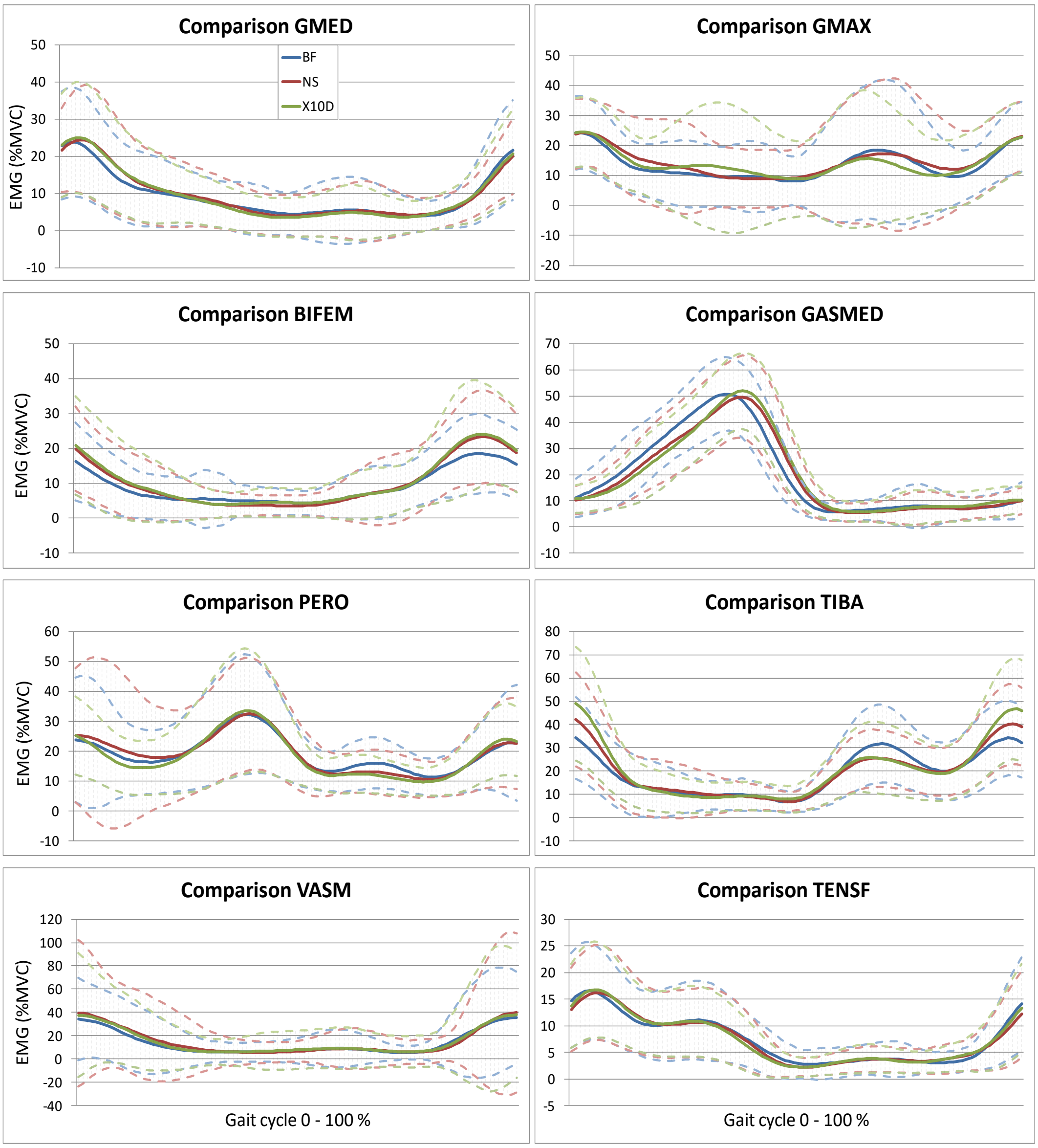


Figure 28 - In each figure the muscle activation over a complete gait cycle for the eight muscles measured is shown. Additionally, the standard deviation is represented by lighter coloured, dashed lines.

5.1.3 Results of time-distance parameters

For the comparison of the time-distance parameters an ANOVA with repeated measures was calculated. The results in Table 7 show a highly significant difference (*p* = 0,000) for the stance phase, the double leg stance phase, the stride length, the stride width, the double stride length, the velocity and the cadence. No differences (*p* = 0,199) were detected for the foot rotation. Concerning the significant p-values further statistics in form of pairwise comparisons (post hoc test) were carried out. The mean of the measured parameters increases from barefoot to norm shoe to the X10D shoe, shown in Figure 29. One exception, however, occurs in the case of cadence, where exactly the opposite is the case. All time-distance parameters, in exception of the foot rotation, differentiate significantly in every possible situation of pairwise comparison.

Time-distance gait parameters

Parameter	BF		NS		X10D			ANOVA	
	Mean	SD	Mean	SD	Mean	SD	df	F	p
Stance phase [%]	64,08	1.74	64,98	1,56	66,23	2,20	2	31,389	**0,000**
DL supp. phase [%]	28,59	3,58	29,72	3,21	32,50	4,50	2	31,700	**0,000**
Stride length [cm]	63,57	4,98	67,85	4,49	68,38	4,65	2	120,907	**0,000**
Stride width [cm]	8,39	2,01	8,57	2,15	9,09	2,06	2	9,022	**0,000**
Dbl. stride l. [cm]	127,06	10,6	135,38	9,41	136,65	10,2	2	113,486	**0,000**
Velocity [km/h]	4,11	0,46	4,28	0,41	4,27	0,49	2	15,584	**0,000**
Cadence [steps/min]	107,99	8,08	105,34	6,95	103,98	7,47	2	29,633	**0,000**
Foot rotation [degree]	11,61	4,39	11,16	3,60	11,00	4,16	2	1,655	0,199

Table 7 - The averaged mean values of the time-distance parameters of all participants with the associated standard deviation for each shoe situation.

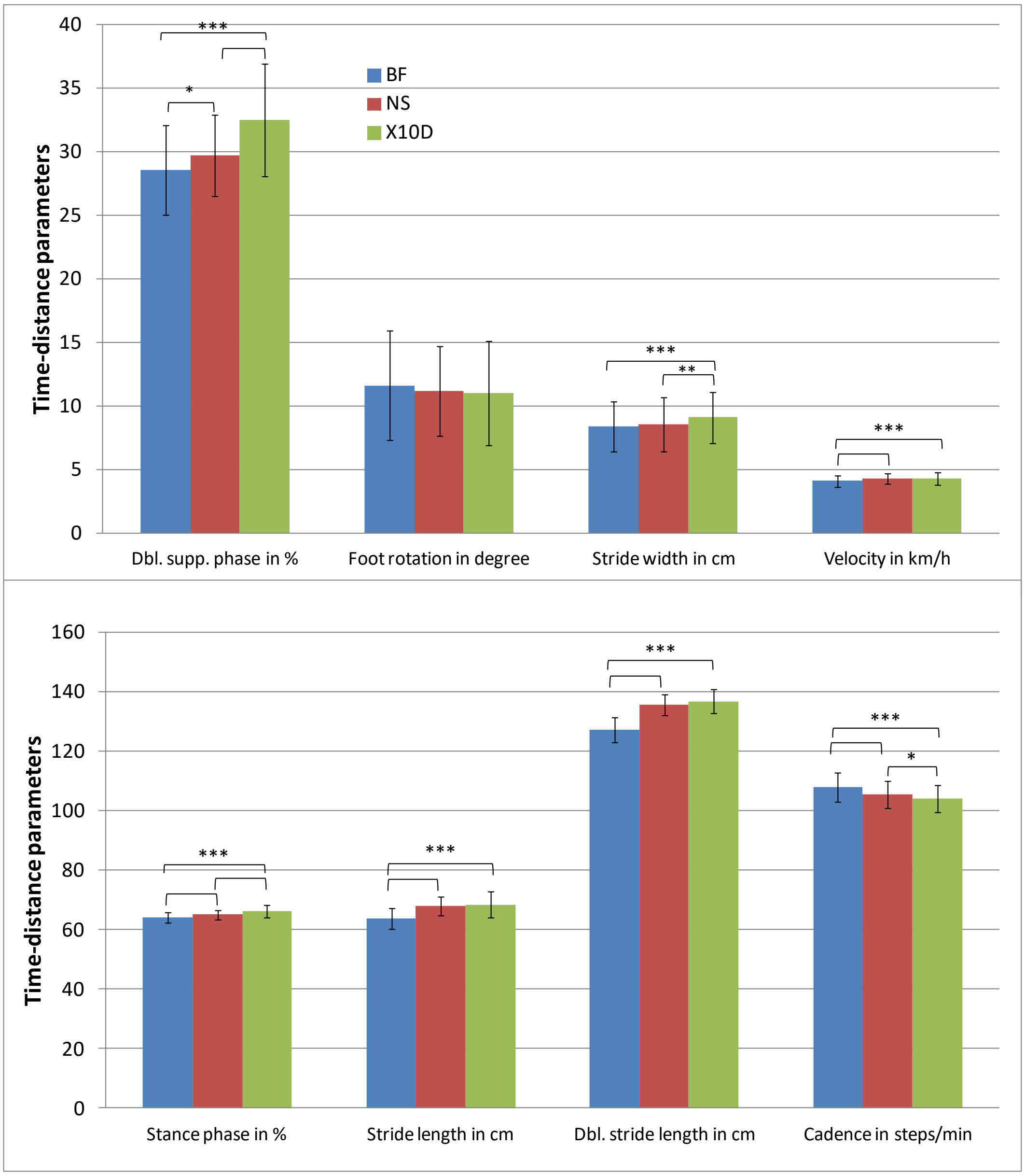


Figure 29 - The averaged values of the time-distance parameters of all participants with the associated standard deviation for each shoe situation.

5.2 Results of the CoP measurements

5.2.1 Results of the double leg stand

For the comparison of the collected data concerning the double leg stance (DL) an ANOVA with repeated measures was calculated. In Table 8 the results show significant differences for the mean distance anterior-posterior (p = 0,043), the root mean square distance anterior-posterior (p = 0,032), the total excursion anterior-posterior (p = 0,001), the mean velocity anterior-posterior (p = 0,001) and the range anterior-posterior (p = 0,017). Concerning the significant p-values further statistics in form of pairwise comparisons (post hoc test) were carried out, which are shown in Figure 30.

Table 8 - The averaged mean values of the double leg standing CoP parameters of all participants with the associated standard deviation for each shoe situation.

CoP double leg									
Parameter	BF		NS		X10D			ANOVA	
[m]	Mean	SD	Mean	SD	Mean	SD	df	F	p
MDISTap	0,0030	0,0008	0,0034	0,0010	0,0034	0,0010	2	3,324	**0,043**
MDISTml	0,0018	0,0007	0,0018	0,0006	0,0019	0,0007	2	1,361	0,264
RDISTap	0,0036	0,0010	0,0040	0,0012	0,0040	0,0012	2	3,637	**0,032**
RDISTml	0,0023	0,0008	0,0022	0,007	0,0024	0,008	2	1,298	0,280
TOTEXap	0,0881	0,0196	0,0956	0,0206	0,0947	0,0195	2	8,401	**0,001**
TOTEXml	0,0737	0,0270	0,0716	0,0223	0,0733	0,0219	2	0,463	0,631
MVELOap	0,0059	0,0013	0,0064	0,0014	0,0063	0,0013	2	8,401	**0,001**
MVELOml	0,0049	0,0018	0,0048	0,0015	0,0049	0,0015	2	0,463	0,631
RANGEap	0,0155	0,0041	0,0174	0,0044	0,0173	0,0043	2	4,382	**0,017**
RANGEml	0,0104	0,0034	0,0101	0,0027	0,0106	0,0035	2	1,035	0,361

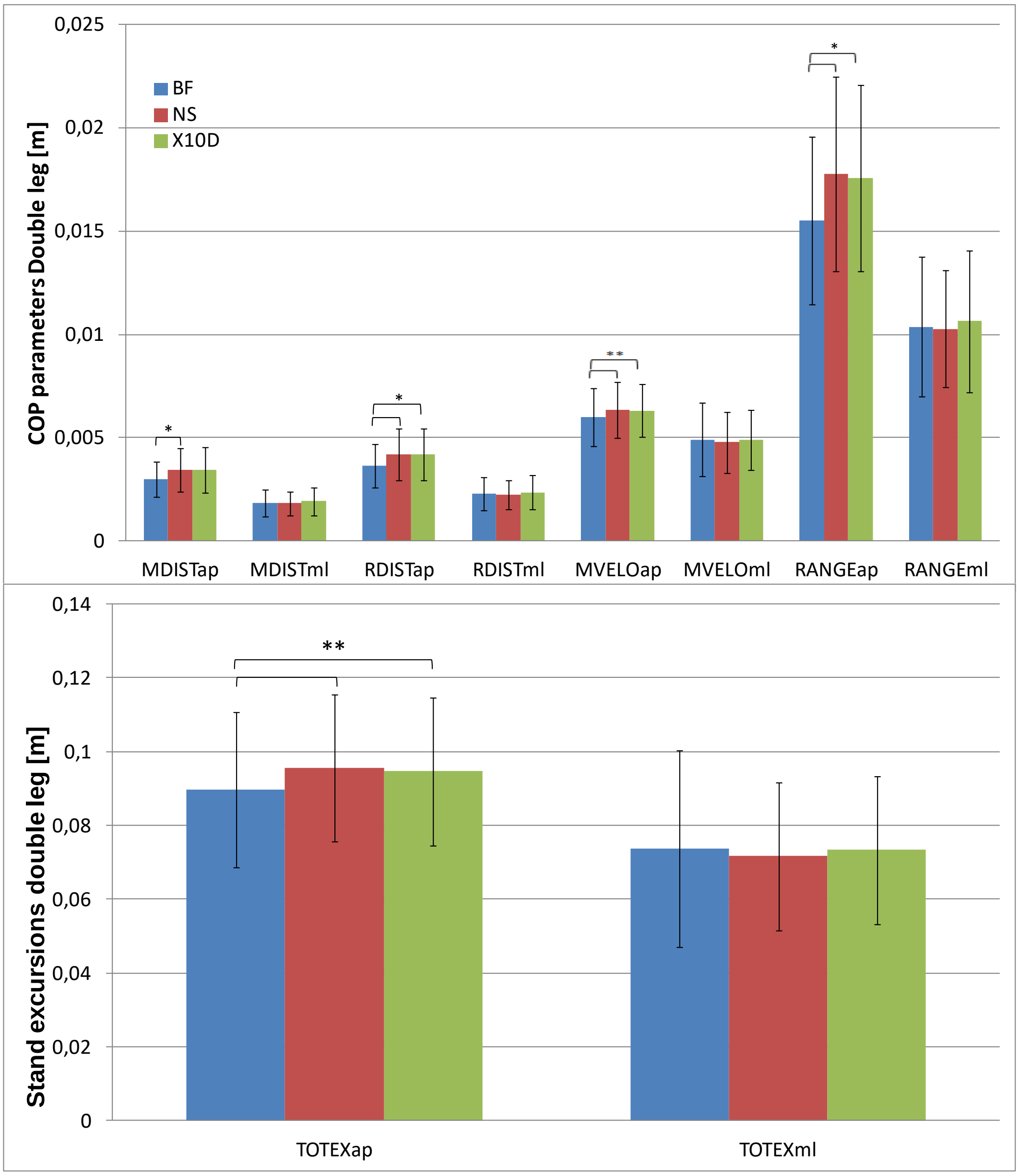


Figure 30 - The averaged values of all participants for the parameters concerning the CoP in double leg standing with the associated standard deviation for each parameter and each shoe situation.

The pairwise comparisons concerning the mean distance anterior-posterior (DL) indicate that only the barefoot situation to the norm shoe (p = 0,032) differentiated significantly. Whereas the norm shoe and the X10D shoe do not differentiate (p = 0,992), the X10D shoe shows a tendency towards significance (p = 0,053) in comparison to the barefoot situation, shown in Table 9. Apart from the pairwise comparisons, the partial eta squared (η^2) for the parameter mentioned above amounts to a value of 0,097.

Table 9 – Pairwise comparisons of the different shoe situations concerning the mean distance anterior-posterior in double leg standing.

Shoe situation		Mean difference [m]	Standard error	Sig.	95% confidence interval for the difference	
					lower limit	upper limit
BF	**NS**	0,000	0,000	0,032	-0,001	-3,5735
	X10D	0,000	0,000	0,053	-0,001	4,696
NS	**BF**	0,000	0,000	0,032	3,573	0,001
	X10D	1,528	0,000	0,992	0,000	0,000
X10D	**BF**	0,000	0,000	0,053	-4,696	0,001
	NS	-1,528	0,000	0,992	0,000	0,000

The pairwise comparisons concerning the root mean square distance anterior-posterior (DL) indicate that the barefoot situation to the norm shoe (p = 0,032) and to the X10D shoe (p = 0,040) differentiate significantly, whereas the norm shoe and the X10D shoe do not differentiate (p = 0,983), which is shown in Table 10. Apart from the pairwise comparisons, the partial eta squared (η^2) for the parameter mentioned above amounts to a value of 0,105.

Table 10 – Pairwise comparisons of the different shoe situations concerning the root mean square distance-anterior posterior in double leg standing.

Shoe situation		Mean difference [m]	Standard error	Sig.	95% confidence interval for the difference	
					lower limit	upper limit
BF	**NS**	0,000	0,000	0,030	-0,001	-4,760
	X10D	0,000	0,000	0,040	-0,001	-2,175
NS	**BF**	0,000	0,000	0,030	4,760	0,001
	X10D	3,641	0,000	0,983	0,000	0,000
X10D	**BF**	0,000	0,000	0,040	2,175	0,001
	NS	-3,641	0,000	0,983	0,000	0,000

The pairwise comparisons concerning the total excursion anterior-posterior (DL) indicate that the barefoot situation to the norm shoe (p = 0,002) and to the X10D shoe (p = 0,002) differentiate highly significant, whereas the norm shoe and the X10D shoe do not differentiate (p = 0,640), which is shown in Table 11. Apart from the pairwise comparisons, the partial eta squared (η^2) for the parameter mentioned above amounts to a value of 0,213.

Table 11 – Pairwise comparisons of the different shoe situations concerning the total excursion anterior-posterior in double leg standing.

Shoe situation		Mean difference [m]	Standard error	Sig.	95% confidence interval for the difference	
					lower limit	upper limit
BF	**NS**	-0,007	0,002	0,002	-0,012	-0,003
	X10D	-0,007	0,002	0,002	-0,010	-0,003
NS	**BF**	0,007	0,002	0,002	0,003	0,012
	X10D	0,001	0,002	0,640	-0,003	0,005
X10D	**BF**	0,007	0,002	0,002	0,003	0,010
	NS	-0,001	0,002	0,640	-0,005	0,003

The pairwise comparisons concerning the mean velocity anterior-posterior (DL) indicate that the barefoot situation to the norm shoe (p = 0,002) and to the X10D shoe (p = 0,002) differentiate highly significant, whereas the norm shoe and the X10D shoe do not differentiate (p = 0,640), which is shown in Table 12. Apart from the pairwise comparisons, the partial eta squared (η^2) for the parameter mentioned above amounts to a value of 0,213.

Table 12 – Pairwise comparisons of the different shoe situations concerning the velocity anterior-posterior in double leg standing.

Shoe situation		Mean difference [m]	Standard error	Sig.	95% confidence interval for the difference	
					lower limit	upper limit
BF	**NS**	0,000	0,000	0,002	-0,001	0,000
	X10D	0,000	0,000	0,002	-0,001	0,000
NS	**BF**	0,000	0,000	0,002	0,000	0,001
	X10D	5,958	0,000	0,640	0,000	0,000
X10D	**BF**	0,000	0,000	0,002	0,000	0,001
	NS	-5,958	0,000	0,640	0,000	0,000

The pairwise comparisons concerning the range anterior-posterior (DL) indicate that the barefoot situation to the norm shoe (*p* = 0,021) and to the X10D shoe (*p* = 0,026) differentiate significantly, whereas the norm shoe and the X10D shoe do not differentiate (*p* = 0,794), which is shown in Table 13. Apart from the pairwise comparisons, the partial eta squared (η^2) for the parameter mentioned above amounts to a value of 0,124.

Table 13 – Pairwise comparisons of the different shoe situations concerning the range anterior-posterior in double leg standing.

Shoe situation		Mean difference [m]	Standard error	Sig.	95% confidence interval for the difference	
					lower limit	upper limit
BF	**NS**	-0,002	0,001	0,021	-0,004	0,000
	X10D	-0,002	0,001	0,026	-0,003	0,000
NS	**BF**	0,002	0,001	0,021	0,000	0,004
	X10D	0,000	0,001	0,794	-0,001	0,001
X10D	**BF**	0,002	0,001	0,026	0,000	0,003
	NS	0,000	0,001	0,794	-0,001	0,001

5.2.2 Results of the single leg stance

For the comparison of the collected data concerning the single leg stance an ANOVA with repeated measures was calculated. In Table 14 the results show significant differences for the mean distance anterior-posterior (p = 0,007), the root mean square distance anterior-posterior (p = 0,012), the total excursion anterior-posterior (p = 0,000), the total excursion medio-lateral (p = 0,006), the mean velocity anterior-posterior (p = 0,000), the mean velocity medio-lateral (p = 0,006) and the range anterior-posterior (p = 0,013). Concerning the significant p-values further statistics in form of pairwise comparisons (post hoc test) were carried out, which are also depicted in Figure 31.

Table 14 - The averaged mean values of the single leg standing CoP parameters of all participants with the associated standard deviation for each shoe situation.

CoP single leg									
Parameter [m]	BF		NS		X10D			ANOVA	
	Mean	SD	Mean	SD	Mean	SD	df	F	p
MDISTap	0,0058	0,0014	0,0065	0,0016	0,0060	0,0012	2	5,295	**0,007**
MDISTml	0,0046	0,0007	0,0046	0,0007	0,0047	0,0007	2	0,620	0,541
RDISTap	0,0072	0,0017	0,0079	0,0018	0,0074	0,0015	2	4,733	**0,012**
RDISTml	0,0056	0,0009	0,0058	0,0008	0,0058	0,0008	2	0,939	0,396
TOTEXap	0,3870	0,0968	0,4011	0,0887	0,4294	0,0973	2	11,903	**0,000**
TOTEXml	0,4280	0,0921	0,4477	0,0937	0,4558	0,0971	2	5,639	**0,006**
MVELOap	0,0253	0,0059	0,0264	0,0056	0,0281	0,0057	2	11,208	**0,000**
MVELOml	0,0285	0,0061	0,0298	0,0062	0,0304	0,0065	2	5,639	**0,006**
RANGEap	0,0346	0,0070	0,0376	0,0071	0,0366	0,0066	2	4,631	**0,013**
RANGEml	0,0269	0,0032	0,0280	0,0032	0,0275	0,0030	2	2,739	0,072

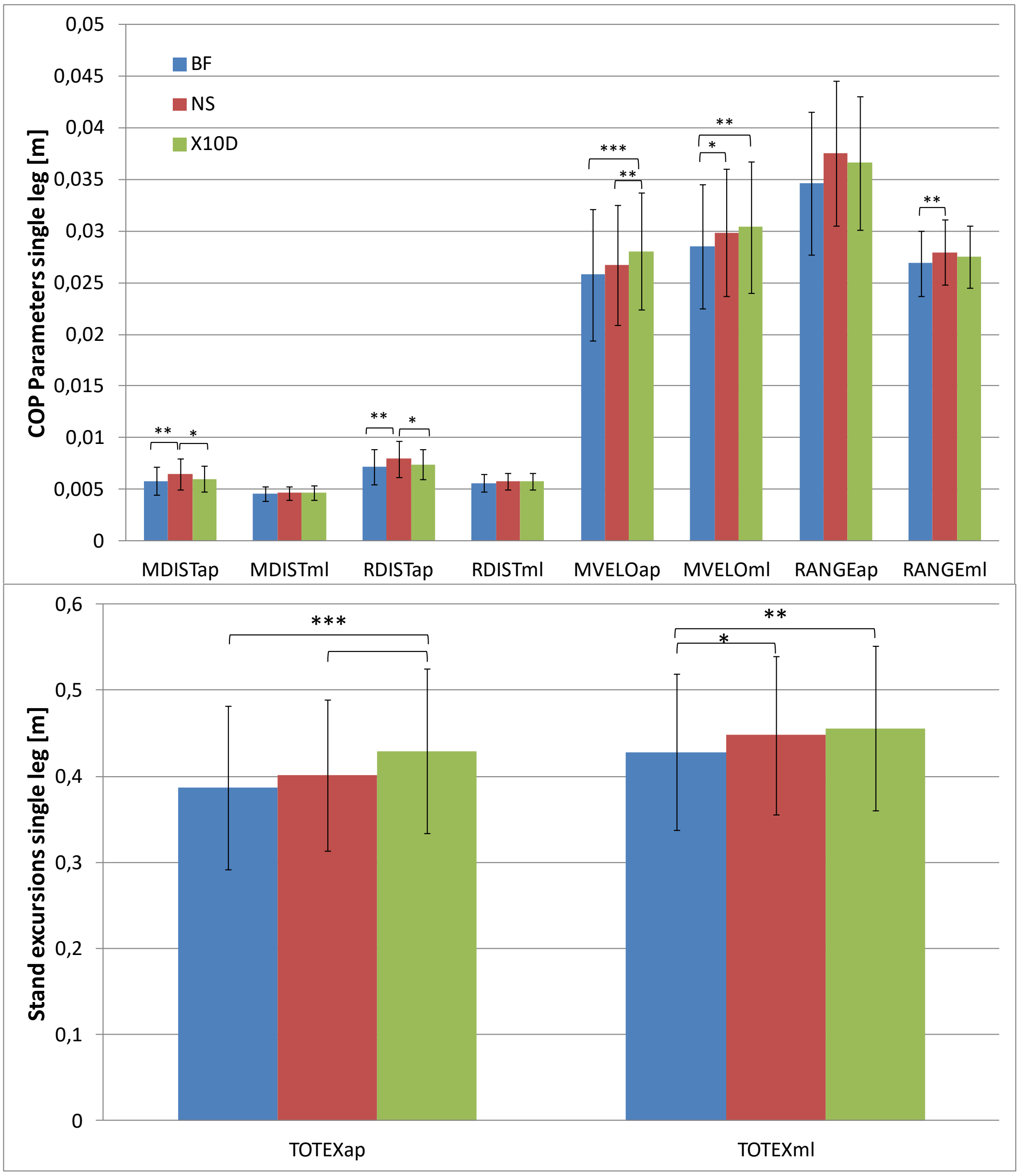


Figure 31 - The averaged values of all participants for some parameters concerning the CoP in single leg standing with the associated standard deviation for each shoe situation.

The pairwise comparisons concerning the mean distance anterior-posterior (SL) indicate that the norm shoe differentiates highly significant to the barefoot situation (p = 0,005) and significant to the X10D shoe (p = 0,038), whereas the X10D shoe does not differentiate significant in comparison to the barefoot situation (p = 0,335), which is shown in Table 15. Apart from the pairwise comparisons, the partial eta squared (η^2) for the parameter mentioned above amounts to a value of 0,142.

Table 15 – Pairwise comparisons of the different shoe situations concerning the mean distance anterior-posterior in single leg standing.

Shoe situation		Mean difference [m]	Standard error	Sig.	95% confidence interval for the difference lower limit	upper limit
BF	**NS**	-0,001	0,000	0,005	-0,001	0,000
	X10D	0,000	0,000	0,335	-0,001	0,000
NS	**BF**	0,001	0,000	0,005	0,000	0,001
	X10D	0,000	0,000	0,038	2,886	0,001
X10D	**BF**	0,000	0,000	0,335	0,000	0,001
	NS	0,000	0,000	0,038	-0,001	-2,886

The pairwise comparisons concerning the root mean square distance anterior-posterior (SL) indicate that the norm shoe differentiates highly significant to the barefoot situation (p = 0,006) and significant to the X10D shoe (p = 0,049), whereas the X10D shoe does not differentiate significant in comparison to the barefoot situation (p = 0,350), which is shown in Table 16. Apart from the pairwise comparisons, the partial eta squared (η^2) for the parameter mentioned above amounts to a value of 0,129.

Table 16 – Pairwise comparisons of the different shoe situations concerning the root mean square distance anterior-posterior in single leg standing.

Shoe situation		Mean difference [m]	Standard error	Sig.	95% confidence interval for the difference lower limit	upper limit
BF	**NS**	-0,001	0,000	0,006	-0,001	0,000
	X10D	0,000	0,000	0,350	-0,001	0,000
NS	**BF**	0,001	0,000	0,006	0,000	0,001
	X10D	0,001	0,000	0,049	1,919	0,001
X10D	**BF**	0,000	0,000	0,350	0,000	0,001
	NS	-0,001	0,000	0,049	-0,001	-1,919

The pairwise comparisons concerning the total excursion anterior-posterior (SL) indicate that the X10D shoe differentiates highly significant to the barefoot situation (p = 0,000) and to the norm shoe (p = 0,001), whereas the norm shoe does not differentiate significant in comparison to the barefoot situation (p = 0,118), which is shown in Table 17. Apart from the pairwise comparisons, the partial eta squared (η^2) for the parameter mentioned above amounts to a value of 0,271.

Table 17 – Pairwise comparisons of the different shoe situations concerning the total excursion anterior-posterior in single leg standing.

Shoe situation		Mean difference [m]	Standard error	Sig.	95% confidence interval for the difference	
					lower limit	upper limit
BF	**NS**	-0,014	0,009	0,118	-0,032	0,004
	X10D	-0,042	0,010	0,000	-0,062	-0,023
NS	**BF**	0,014	0,009	0,118	-0,004	0,032
	X10D	-0,028	0,008	0,001	-0,044	-0,012
X10D	**BF**	0,042	0,010	0,000	0,023	0,062
	NS	0,028	0,008	0,001	0,012	0,044

The pairwise comparisons concerning the total excursion medio-lateral (SL) indicate that the barefoot situation to the norm shoe differentiate significantly (p = 0,025) and to the X10D shoe even highly significant (p = 0,003), whereas the norm shoe and the X10D shoe do not differentiate significant (p = 0,345), which is shown in Table 18. Apart from the pairwise comparisons, the partial eta squared (η^2) for the parameter mentioned above amounts to a value of 0,150.

Table 18 – Pairwise comparisons of the different shoe situations concerning the total excursion medio-lateral in single leg standing.

Shoe situation		Mean difference [m]	Standard error	Sig.	95% confidence interval for the difference lower limit	upper limit
BF	**NS**	-0,020	0,008	0,025	-0,037	-0,003
	X10D	-0,028	0,009	0,003	-0,045	-0,010
NS	**BF**	0,020	0,008	0,025	0,003	0,037
	X10D	-0,008	0,008	0,345	-0,025	0,009
X10D	**BF**	0,028	0,009	0,003	0,010	0,045
	NS	0,008	0,008	0,345	-0,009	0,025

The pairwise comparisons concerning the mean velocity anterior-posterior (SL) indicate that the X10D shoe differentiates highly significant to the barefoot situation (p = 0,000) and to the norm shoe (p = 0,002), whereas the norm shoe in comparison to the barefoot situation only shows a tendency towards significance (p = 0,074), which is shown in Table 19. Apart from the pairwise comparisons, the partial eta squared (η^2) for the parameter mentioned above amounts to a value of 0,266.

Table 19 – Pairwise comparisons of the different shoe situations concerning the mean velocity anterior-posterior in single leg standing.

Shoe situation		Mean difference [m]	Standard error	Sig.	95% confidence interval for the difference lower limit	upper limit
BF	**NS**	-0,001	0,001	0,074	-0,002	0,000
	X10D	-0,003	0,001	0,000	-0,004	-0,001
NS	**BF**	0,001	0,001	0,074	0,000	0,002
	X10D	-0,002	0,001	0,002	-0,003	-0,001
X10D	**BF**	0,003	0,001	0,000	0,001	0,004
	NS	0,002	0,001	0,002	0,001	0,003

The pairwise comparisons concerning the mean velocity medio-lateral (SL) indicate that the barefoot situation to the norm shoe differentiate significantly (*p* = 0,025) and to the X10D shoe even highly significant (*p* = 0,003), whereas the norm shoe and the X10D shoe do not differentiate significantly (*p* = 0,345), which is shown in Table 20. Apart from the pairwise comparisons, the partial eta squared (η^2) for the parameter mentioned above amounts to a value of 0,150.

Table 20 – Pairwise comparisons of the different shoe situations concerning the mean velocity medio-lateral in single leg standing.

Shoe situation		Mean difference [m]	Standard error	Sig.	95% confidence interval for the difference	
					lower limit	upper limit
BF	**NS**	-0,001	0,001	0,025	-0,002	0,000
	X10D	-0,002	0,001	0,003	-0,003	-0,001
NS	**BF**	0,001	0,001	0,025	0,000	0,002
	X10D	-0,001	0,001	0,345	-0,002	0,001
X10D	**BF**	0,002	0,001	0,003	0,001	0,003
	NS	0,001	0,001	0,345	-0,001	0,002

The pairwise comparisons concerning the range anterior-posterior (SL) indicate that the barefoot situation to the norm shoe differentiate highly significant (*p* = 0,004), whereas in comparison to the X10D only a tendency towards significance (*p* = 0,055) is given. The norm shoe and the X10D shoe do not differentiate significantly (*p* = 0,346), which is shown in Table 21. Apart from the pairwise comparisons, the partial eta squared (η^2)for the parameter mentioned above amounts to a value of 0,126.

Table 21 – Pairwise comparisons of the different shoe situations concerning the range anterior-posterior in single leg standing.

Shoe situation		Mean difference [m]	Standard error	Sig.	95% confidence interval for the difference	
					lower limit	upper limit
BF	**NS**	-0,003	0,001	0,004	-0,005	-0,001
	X10D	-0,002	0,001	0,055	-0,004	4,317
NS	**BF**	0,003	0,001	0,004	0,001	0,005
	X10D	0,001	0,001	0,346	-0,001	0,003
X10D	**BF**	0,002	0,001	0,055	-4,317	0,004
	NS	-0,001	0,001	0,346	-0,003	0,001

For the range medio-lateral no pairwise comparisons were carried out, because the results only show a tendency towards significance (p = 0,072).

5.3 Subjective feeling and feedback

The collections of the subjective feeling and feedback regarding the look, the fit, the materials used, the weight, the workmanship and the first impression were divided into three categories:

- Properties of the shoes
- First impressions when walking
- First impressions when standing

Regarding the properties of the X10D many of the participants had the impression that the shoe is less breathable and therefore too warm or sweaty. The material looks good and easy to care for most of the participants and the proper lacing was highlighted. Some reported that the shoe has a fairly hard cushioning, while others say the shoe generally feels a bit stiff and firm. Opinions differed widely regarding the design of the shoe. While the older participants were quite convinced of the simple leather design, many of the younger participants would not wear this shoe in their spare time due to the design.

Concerning the first impression when wearing the shoe, opinions did not differ that much. Most participants reported an unfamiliar feeling when walking with the

X10D. Some of the subjects also reported having the feeling of tilting inwards more often.

During double leg stance in turn, most subjects had a very good and safe feeling due to the flat shoe sole, the robust leather upper material and the solid hold in the shoe. However, during the one leg stance things looked very different. Almost all participants mentioned that the shoe felt particularly shaky and described difficulties by pushing the body weight over the lateral edge of the foot.

6 Discussion

In the following chapter, the results of the statistical data analysis will be discussed, interpreted and compared with the current state of science in order to clarify the research questions and research hypotheses of this study. The purpose of this thesis was to evaluate the influence of a new unstable footwear construction on muscle activity and postural control. One of the main objectives of this study was to clarify if and how the initial wearing of the X10D shoe affects the maximum and average muscle activity of the m. gluteus medius, m. gluteus maximus, m. tensor fasciae latae, m. quadriceps femoris vastus medialis, m. gastrocnemius medialis, m. tibialis anterior and m. peroneus longus during walking in adults between 18 and 65 years of age. The other main objective was to clarify if and what impact wearing the X10D shoe has on the postural control and thus on the CoP excursions. This was examined while standing on a force plate, in two different positions (double leg and single leg support). From this context, the following hypotheses emerged:

- *The first hypothesis was, that there is a significant difference in maximum muscle activity of lower extremity muscles when wearing the X10D shoe compared to a reference shoe and barefoot.*
- *The second hypothesis was, that there is a significant difference in average muscle activity of lower extremity muscles when wearing the X10D shoe compared to a reference shoe and barefoot.*

Regarding the CoP fluctuations, the hypotheses were the following:

- *Wearing the X10D shoe does affect the fluctuations of postural control significantly when standing upright compared to a reference shoe and barefoot.*
- *Wearing the X10D shoe does affect the fluctuations of postural control significantly when standing upright on one leg compared to a reference shoe and barefoot.*

6.1 Interpretation of the results

To test the first research hypothesis, an ANOVA with repeated measures was calculated for the respective muscles under the different measurement conditions - barefoot, norm shoe and X10D. Two of the eight muscles indicate a significant difference in EMG peak values. The results for the peak muscle activity of m. gluteus medius, m. gluteus maximus, m. gastrocnemius caput medialis, m. peroneus longus, m. quadriceps femoris vastus medialis and m. tensor fasciae latae were not significant. Therefore, the first research hypothesis, which states that a significant difference in maximum muscle activity between the shoe conditions can be identified, must be rejected for these muscles. Only the evaluations of m. tibialis anterior (p = 0,002) and m. biceps femoris (p = 0,047) showed a different result. The maximum muscle activity of m. tibialis anterior showed a highly significant difference between barefoot and the X10D. In case of m. biceps femoris no significant differences could be measured in the pairwise comparisons between the three testing situations. Therefore, the first research hypothesis may be accepted for the m. tibialis anterior - there is a difference in maximum muscle activity when first wearing X10D shoe compared to barefoot walking and walking with a reference shoe.

The second research hypothesis regarding mean muscle activity must be rejected for all measured muscles. None of the eight muscles measured showed a difference in mean muscle activity between the three shoe situations.

The third research hypothesis, which focused on the CoP excursions during double leg stance, could not be confirmed for the measured parameters in medio-lateral direction. Nevertheless, MDISTap, RDISTap, TOTEXap; MVELOap and RANGEap showed a significant difference. But in pairwise comparisons only significant differences between barefoot and norm shoe or barefoot and the X10D shoe was found. No significant differences could be detected between the norm shoe and the X10D. Therefore, the third research hypothesis that wearing the X10D shoe does affect the fluctuations of postural control significantly when standing upright compared to a reference shoe and barefoot, can only be partly confirmed for the parameters in the anterior-posterior direction.

The fourth research hypothesis, which focused on the CoP excursions during single leg stance, could not be confirmed for MDISTml, RDISTml and RANGEml, because no significant differences could be detected after carrying out an ANOVA with repeated measures. Nevertheless, significant differences appeared among all the other parameters (MDISTap, RDISTap, TOTEXap, TOTEXml, MVELOap, MVELOml, RANGEap). However, the fourth research hypothesis that wearing the X10D shoe does affect the fluctuations of postural control significantly when standing upright on one leg compared to a reference shoe and barefoot, can only be confirmed partly for MDISTap, RDISTap, TOTEXml, MVELOml and RANGEap. In each of these cases, a difference between the situations could not be established in all pairwise comparisons. For TOTEXap and MVELOap, however, the hypothesis can be considered valid.

The tibialis anterior muscle, as already described in the theoretical background, is mainly eccentrically active in the stance phase and, thus, controls the lowering of the foot. In addition, it is responsible for the pronation of the lower ankle joint, together with the m. posterior tibialis. One possible explanation for the significantly higher tibialis anterior EMG levels in the stance phase during the measurements is due to the height of the X10D shoe soles. The sole thickness of the X10D is 3.5 cm at the heel, while barefoot did not increase the heel. The mean sole thickness of the reference shoes was, in our estimation, between these two values, but was not explicitly measured. The increased maximum activity of m. tibialis anterior could result from the different sole thickness and the resulting different leverage effects. This could affect the heel strike in initial contact and the subsequent lowering of the foot. This theory can be supported by an analysis of the EMG curves of the gait reports (Figure 28). The EMG curves show increased muscle activity of m. tibialis anterior, especially at the beginning of the stance phase. Then the curve drops flat, almost to the zero line. An increased activity is thereafter not visible until the end of stance phase. This theory can also be reconciled with other studies. Nigg et al., (2006) describes an increased dorsiflexion of the ankle at the beginning of the stance phase. In this context, however, it must be mentioned that the investigations by Nigg et al., (2006) were executed on a different shoe (MBT shoe).

Another aspect that determines the maximum activity of the m. tibialis anterior, is the weight of the different shoes, in alternating test situations. The X10D was distinctly the heaviest at an average weight of 363 ± 68 g per shoe. In comparison, the average weight of the norm shoes was 260 ± 34 g per shoe. These weight differences could increase the activity of the m. tibialis anterior at the beginning of the stance phase. The reason for this is the additional weight, which has to be counteracted by increased muscular activity. This theory may be reflected in the measurement results.

Another possible explanation for the significant increase in muscle activity of the m. tibialis anterior could be the special sole of the X10D shoe. Due to the lack of medial stabilization of the sole, the foot tends to tilt inwards. To correct that, it is necessary to raise the arch of the foot through muscle activation. Primarily this task is carried out by the m. tibialis posterior but the m. tibialis anterior is supporting this process. Landry, Nigg, & Tecante (2010) argue that many of the smaller extrinsic foot muscles, such as the m. tibialis anterior, have a favourable anatomical position for the control of lower ankle movements in order to be able to change their position more rapidly. This suggests that muscles such as the m. tibialis anterior effectively affect the balance. The muscle activity of m. tibialis anterior was increased especially during initial contact and loading response. During this phase, the anterior tibial musculature lowers the forefoot and supports the stabilization of the ankle joint. Lack of stability caused by the shoe would therefore result in increased muscular activity.

The fact that no significant differences in the mean muscle activity during the different shoe conditions were measurable, leads to different explanations. A fairly simple explanation would be, that the measured muscules were not the muscles recruited by the subjects to compensate the instability caused by the X10D. According to Papalia et al. (2015), balance strategies vary according to the individual anatomy and body constitution of the participants. In order to maintain balance, strategies that use the ankle as a stabilizer are therefore applied. However, often a mixture of various strategies, including muscles around the hip or knee, are often used to avoid being unbalanced. The ankle strategy, for example, involves muscle activation of musculature that pulls over the upper ankle joint, while the hip strategy activates hip and trunk muscles first to compensate for instability (Knuchel & Schädler, 2004). Both strategies of balance control were picked up in this study. However, neither in the hip muscles nor in those around the ankle effects could be detected. The deep calf muscles also influence the arch of the foot and the stability in the ankle joint (Platzer, 2009, pp. 262-265). However, the activity of these muscles cannot be detected with a surface EMG. Another explanatory approach could be that these muscles compensate the instability caused by the X10D. However, these muscles were not recorded during the measurements.

Price, Smith, Graham-Smith, & Jones, (2013) criticize that recorded data is normalized to an MVC measurement, as it was carried out in this study. Disadvantages of this method are among other things, the influence by the daily constitution of the person to be tested. Fatigue, pain, posture, performance and motivation of the participant can have a strong influence on the measurement results (Konrad, 2005). In addition to that, Konrad (2005) claims that static muscle testing can be considered problematic, because the muscle length in these tests may differ from the muscle length in dynamic motions under investigation and therefore the tests are not representative. Nevertheless, in recent studies MVC measurement is the most common method of amplitude normalization.

To determine the pure effects of the X10D on muscle activity, subjects did not have time to become familiar with the shoe. This differs from the studies of Horsak & Baca (2013), Landry et al. (2010) and Romkes (2008), where a period of adjustment was given. However, this study deliberately avoided a period of adjustment to exclude clenched up muscles and potential changes in muscle activity through conscious walking, as described by Wulf (2009).

The results of this study correlate very strongly with those of Branthwaite, Chockalingam, Pandyan, & Khatri (2013). They did not notice a short-term effect of an unstable sole construction (MBT shoe) compared to a sports shoe. Rather, they describe very individual EMG results of the subjects. Just as in this study, the test persons were not familiarized in advance with the handling of the MBT shoe.

Although only minor effects of the X10D could be recorded while walking, many of the subjects reported that they had an unfamiliar feeling while walking. Many participants described the feeling of walking incresingly on the lateral edge of the foot. It can be concluded that the subjects need a little more time to get used to walking with the X10D. The investigations of Swager van Dak, Baur, Cabri, & Hirschmüller (2015) have deposited this. They reported, that after eight weeks of accommodation to the X10D, significantly higher peak pressures were measured for the lateral midfoot with the X10D and barefoot compared to a reference shoe.

For the significant differences of the time-distance parameters between the three different situations (barefoot, norm shoe, X10D) there are several explanatory approaches. One explanation for this is the damping of the shoes at heel strike. While the sole of the shoes provide some cushioning for the impact on the ground, this extra cushioning is not present when walking barefoot. Furthermore, the sole thickness obviously has an influence on the time-distance parameters, since these parameters increased with the thickness of the sole. All time-distance parameters, apart from the cadence, are lowest for barefoot and highest for the X10D. As already mentioned above, the X10D had a much thicker sole than the reference shoes. The differences of the time-distance parameters can therefore also be explained by the fact that the initial contact is triggered earlier by the "wedge" under the heel. Furthermore, the length of the shoe sole also influences the measured stride length, as the heel strike takes place earlier and the toe off later, in comparison to walking barefoot. All these facts thus indicate a systematic measurement error, which arises due to the different shoe models. However, these minor deviations should not affect the definition of the gait phases by the

Noraxon software. In addition, an external system with two light barriers and not the pressure measuring plate, was used to ensure a constant walking speed Nevertheless, Demura, Demura, & Yamada (2012) described similar results in their study. They point out that shoes may affect walking speed and speed may depend on stride length and cadence. Furthermore, they point out that the walking pace with shoes is increased by the increased stride length and the quality of the shoe material. As in the studies of Horsak & Baca (2013) and Demura et al. (2012), we also compared the unstable shoes with the participants own shoes. Farzadi, Nemati, Jalali, Doulagh, & Kamali (2017), reported participants familiarity with footwear would potentially have impact on gait parameters, especially on speed.

In contrast to several other studies (Farzadi et al., 2017; Nigg et al., 2006 and Stöggl, Haudum, Birklbauer, Murrer, & Müller, 2010), which recorded an increase in CoP excursions in medio-lateral and anterior-posterior direction during double leg stance by using unstable shoes, this was not the case in this study. Although there were some significant differences for several parameters in anterior-posterior direction between barefoot and the X10D, the same differences occurred between barefoot and the reference shoe. So, this would only confirm that there is a difference in CoP excursions between the barefoot situation and any shoe. As a result, the X10D has not shown to affect postural control in standing. This is also in line with the opinions of the participants, who had a good and safe feeling while standing on the force plate with the X10D, in the double-leg scenario.

These results seem to be in contrast with the findings of Nigg et al. (2006). They reported significantly greater CoP excursion in anterior-posterior and medio-lateral direction when standing in MBT shoes. Therefore, they concluded that this shoe concept (MBT) may serve as an effective training device for muscle strength, stability and proprioception. However, it must be said that both shoe concepts differ significantly in the way their soles are constructed. Therefore a direct comparison may not be appropriate. Even though, no increased instability was detectable during quiet standing, the specific construction of the X10D shoe, may only induce instability during walking, especially during loading response and early mid stance. However, the fact that there were no significant differences during the double-leg stance could be due to the composition of the shoe. The sole of the X10D is flat and stiff, the outer material is solid leather and the lacing is very compact. All these factors can provide stability and influence the CoP excursions in stance. In addition, the risk of tilting inwards is not present, since the second leg supports and counteracts.

Other results were found in the evaluation of CoP excursions in the single leg stance scenario. Due to the smaller base of support, single leg standing is a more challenging postural position compared to double leg standing. Therefore, CoP excursions where greater than during double leg standing. TOTEXap and MVELOap showed significant differences. It can be concluded that the X10D may have an influence on CoP excursions in anterior posterior direction, which is a sign for increased instability. An explanation for this would be that the sole construction of the X10D increases the plantar pressure distribution over the lateral margin of the foot, because the shoe provides support here. Based on its construction, the X10D shoe could have forced the user to stand more towards the midfoot. This can be provided by raising up the arch of the foot. The muscles responsible for this task are also responsible for plantarflexion and dorsiflexion in the ankle and so consequently, the anterior-posterior movement was different. These discoveries can also be reconciled with the results of Swager van Dak et al. (2015), who found an increase in plantar pressure distribution due to the wearing of the X10D, especially in the midfoot area.

The fact that no significant differences, in favour of the X10D, in medio-lateral direction could be detected, could be attributed to different standing postures. In this study the subjects were encouraged to lift the free leg and bend hip and knee to 90°. To ensure this, the center of gravity shifts to the other leg directly over the support surface and the pressure load on the outer edge of the foot increases. This results in a certain basic tension of the hip musculature and the participants can, therefore, possibly better compensate for CoP fluctuations. In the study of Plom et al. (2014), which provided a similar standing posture during the tests, only differences in the anterior-posterior direction could be detected. However, Romkes (2008) achieved similar results although her subjects lifted the free leg only a few centimetres off the ground. Again, it should be noted that these two studies investigated other unstable shoes.

The specific sole construction of the X10D could also influence the measurements. The X10D lacks the medial part of the shoe sole, which could cause the forces to be redirected towards the big toe or heel and thus be more likely to be associated with anterior posterior variations. However, this is only a theory that emerged during the implementation of this study and therefore cannot be confirmed by the investigations of others.

Furthermore, the results can always be dependent on the testing situation itself. Turbanski et al. (2011) investigated two different unstable shoes in two different testing situations and concluded that detection of training effects on balance control depends on the testing situation.

Generally, comparisons with other unstable shoes are difficult. Shoe constructions differ significantly from each other and therefore often lead to different results under investigation. Price et al. (2013) also concluded, that the effects of unstable shoe constructions are very product specific.

6.2 Clinical relevance

Even though it seems like the X10D shoe and its special sole construction affects the activation of the m. biceps femoris and the m. tibialis anterior the overall effects were relatively low. The muscle activity of the m. biceps femoris changed by 0.67% (X10D to NS) and 0.44% (X10D to BF). The muscle activity of the m. tibialis anterior changed by 1,57% (X10D to NS) and 3,09% (X10D to BF). Therefore, it cannot be assumed that the shoe affects the musculature of the lower extremity, in the sense of an increase in strength.

In contrast, the results of the stance analysis may indicate a therapeutic relevance. During the one leg stance, significant differences in COP excursion in the anterior posterior direction ocurred. TOTEXap increased by 7,1% (X10D to NS) and 11,0% (X10D to BF). MVELOap increased by 6,4% (X10D to NS) and 11,1% (X10D to BF). Increased CoP excursions indicate a higher requirement of the postural control system. As mentioned in the introduction, the body relies on a neuromuscular interaction to maintain balance. The results could be an indication that the X10D influences this neuromuscular interaction, in the sense of a sensorimotor training. Therefore, the X10D could be seen as a supporting tool for a sensorimotor training in therapeutical settings.

Korsten, Mornieux, Walter & Gollhofer (2008) see unstable sole constructions as an alternative to conventional sensorimotor training on the wobble board or wobble mats. Nevertheless, it must be mentioned that the adaptation effects are less when training with unstable sole constructions and, therefore, represent a lighter version of sensorimotor training. Thus, the X10D shoe should not be seen as a substitute for conventional sensorimotor training, but rather as an accompanying supportive measure for everyday life, if used properly. By improving the promotion of perception, wearing the X10D shoe could have some positive effects on malpositions of the lower extremities, such as overpronation or knee valgus.

In summary, the X10D could be useful in the prevention as well as in compensating initial stages of lower extremities malpositions. We would like to point out that it is very important that the wearer is adequately informed and instructed on the correct use of the shoe. Ideally, this instruction is carried out by a physical therapist, who can control the execution and consequently is able to provide advice, so that a deterioration of existing misalignments is excluded.

6.3 Limitations

In this chapter some limitations, which could have influenced the study results, are explained in detail. In the course of carrying out this study and processing the data, we were confronted to some limitations that arose due to a lack of resources.

To keep the sample size as high as possible, participants were not excluded by a certain body mass index (BMI). In addition, we wanted to be able to draw better conclusions about the total population. During the execution of the measurements and especially in data evaluation, problems arose with regard to this topic. In the data processing it was recognized that certain participants rather tented to have several outliers in the EMG signal than others. In almost all of the cases those outliers were produced by the overweight participants. In this study 7 of the 33 selected participants had a BMI between 25 and 29,9 which is categorized as overweight (Flegal et al., 2012). The assumption is that an increase in body mass could lead to a higher movement of the electrodes which could influence the quality of the EMG signals. Buchecker, Wagner, Pfusterschmied, Stöggl, & Müller (2012) support this statement with their claim that inaccuracies in the EMG data recording are caused by a higher skin movement especially from overweight people. They add that too little research was done on the influence of overweight and the impact on the EMG recording. Thus, we would recommend, to take an exclusion of participants with a BMI over 25, as where the start of overweight is defined (Flegal, Carroll, Kit, & Ogden, 2012), in consideration.

Another limitation concerning the EMG measuring is the amplitude normalization by the support of the maximum voluntary contraction (MVC) value. Nevertheless, this method is the most common in studies of this kind it is still error prone and controversially discussed in the literature (Konrad, 2005; Price et al., 2013). In this method the maximum contraction of each muscle is used to standardize the EMG values while walking. The amplitude data is then expressed as a percentage of the reference value of the maximum contraction. The MVC value itself can be influenced by several factors. On the one side it can depend on the participants themselves and their daily constitution, their motivation and occurring pain. On the other side the rater has an influence as well. The rater is responsible for the correct starting position and giving a reasonable resistance against the participants muscle contraction. Furthermore, the MVC value only reflects muscle activity in an isometric contraction, this may differ from muscle activity during dynamic movements. Under certain circumstances the recorded data could be falsified and some significant differences in the statistical analysis could not be shown (Sousa & Tavares, 2012).

The subjects of this study were graded by a clinical examination of the arch of the foot and the leg axis. These two points have been rated by visual inspection of two physiotherapists and were categorized on a scale from -5 to +5. Whereas -5 stands for valgus in the knee joint respectively pronation in the ankle joint and +5 for varus in the knee joint respectively supination in the ankle joint. The problem with this method is that it was done on the basis of subjective assessment. For that reason, the classification was not included in the statistical analysis. Furthermore, the number of participants in the subgroups was quite small. For example, with the classification of the arch in terms of the Arch Index or Foot Posture Index (Redmond, Crosbie, & Ouvrier, 2006), a more accurate classification could have been made in order to form homogeneous groups with regard to the arch of the foot. Additionally, the leg axis could have been measured using a goniometer to obtain objective results. In terms of the number of subgroups, it is recommended to keep the subgroups smaller in order to increase the number of participants in a group.

Controversial to other studies, where participants had a period of adjustment (Horsak & Baca, 2013), the X10D shoe was neither explained to the participants, nor they were able to familiarize with the shoe. Recent literature indicates that such an explanation or familiarization could lead to different results. For example, Nigg et al. (2006) as well as Taniguchi, Tateuchi, Takeoka, & Ichihashi (2012) claim that the MBT shoe activates the small muscle groups around the ankle, in comparison to a reference shoe. Landry et al. (2010) even found an increased activation of the flexor digitorum longus muscle and the anterior compartment musculature when wearing the MBT shoe. In these studies, the subjects were informed about the handling of this shoe in advance. In contrast to these studies, Branthwaite et al. (2013) decided to not inform the participants about the handling of the MBT shoe. They thought that the results then could be applied to everyday life, because an enrolment for the wearer would not be necessary. However, they did not notice an effect on the MBT shoe compared to a reference shoe. The results support the claim that the deliberate and correct use of an unstable shoe construction can lead to different outcomes. Another point where the study of Branthwaite et al. (2013) and this study regarding the X10D correlate, is the selection of the reference shoe. Both of them decided that the reference shoe is brought by the study subjects themselves and is not chosen from the authors as in the studies of Landry et al. (2010); Nigg, Hintzen, & Ferber (2006) and Taniguchi et al. (2012). In this case, one must decide whether the results should be assumed as generally valid and additionally considering that this factor may also affect the results.

Another limitation relates to the statistical evaluation, in particular to the alpha error cumulation concerning the pairwise comparisons. When the analysis of variances (ANOVA) becomes significant the post hoc tests in form of pairwise comparisons between the groups are of high interest in order to find out exactly how they differ. Statistical programs provide a variety of tests, whereas only some of them will be discussed here. All of these tests are multiple, a single null hypothesis is examined with multiple tests. In the case of multiple testing there is to consider that the first type of error (alpha error) must be adjusted by means of an error correction. However, there is the strictness of the test, which needs to be taken in consideration, as well as the robustness of the test, for example against normal distribution violations or unequal variances. Some of the tests, such as Bonferroni and Turkey are very strict against the alpha error, which consequences in a low test strength and is therefore determined as too conservative for this study. In contrast, tests like LSD (least significant difference) and S-N-K (studentized Newman-Keuls) have no control of the alpha error and are therefore often determined as too liberal. Since this study is an exploratory study, it has been decided to apply the LSD correction. Thus, it should be noted that the results of the pairwise comparisons tend to indicate significant differences, even though they might would have had to be discarded. This conscious decision was made, in terms of preferring to risk an alpha error instead of ignoring an effect of the shoes and thus commit a beta error.

7 Conclusion

In the last chapter we summarize the whole topic of this thesis, present the conclusion of this study and provide an outlook on future studies.

Walking is an integral part of human life and a very complex neuromuscular activity. Postural control as well as stability are two major factors to ensure natural gait. Shoes are increasingly changing our gait and thus favouring the development of lower extremity pathologies. This in turn leads to further postural damage. Since this has been known for some time, new shoe constructions have been developed which imitate barefoot walking and thus counteract these changes. The actual effect of these sole constructions and their benefit for the wearer have been discussed in numerous studies. In this study, a fairly new shoe, the X10D was examined. This shoe is characterized mainly by the absence of the medial part of the sole.

Regarding muscle activity, significant differences occurred in maximal activation of the m. tibialis anterior. A look at the EMG data reveals that this difference has occurred during initial contact and loading response. The increased muscle activity could have been caused by the special shape of the sole. Instability initiated by the sole could encourage the wearer to a higher muscle activity to maintain balance. However, all other muscles measured showed no significant differences in either maximal or mean muscle activity.

With regard to the CoP excursions, no significant differences, in favour of the X10D shoe, were found during double leg stance. At one leg stance, two of the measured parameters (TOTEXap, MVELOap) differed significantly from the reference shoe and the barefoot situation. It may therefore be suspected that wearing the X10D has an impact on anterior-posterior CoP excursion. This could also be due to the special sole construction.

This study was able to provide initial results and conclusions regarding the effects of the X10D on muscle activity during walking and posture control in standing. However, as the differences were small and the majority of the measured parameters showed no significant differences, the clinical relevance of the observed effects is questionable. In order to be able to make more precise statements regarding the clinical relevance further studies would be necessary.

A future approach to study the effects of this shoe could be an analysis of lower extremity kinetics and kinematics during walking. This would allow more accurate conclusions about the load and the forces acting on the joints. As a result, it may also be possible to recognize connections to the present results. Furthermore, an analysis with a motion capture system could evaluate whether the X10D has a positive influence on the arch of the foot or the leg axis.

Due to the claims in literature that unstable footwear could have an influence on lower back pain (Armand et al., 2014), it would also be interesting to explore if there is a difference between the shoe situations in back or abdominal muscle activation. Another approach could be to compare the dominant and none dominant leg, to see if there is a difference due to a different sensory innervation. Some could expect to detect bigger differences in CoP excursion on the none dominant leg, because it could be harder to compensate the lack of stability of the shoe.

In further consequence it would be extremely interesting, whether long-term effects regarding the muscle activity and the postural control can be determined. In addition, it is important to clarify whether certain effects may occur only after a certain period of familiarization and training. This approach would be verifiable through an intervention study with a correspondingly long intervention period. It would be possible that any changes in movement, posture and muscle activity can be measured only after four to six weeks, when the muscles had enough time to adapt. Since manufacturers emphasize the positive influence on the pressure distribution in the foot and the erection of the arch of the foot, it would be an advantage in subsequent studies to include the evaluation of the arch of the foot. The classification of the arch of the foot with the Foot Posture Index, a reliable diagnostic tool based on six criteria, would be conceivable (Redmond et al., 2006).

To give better recommendations in terms of therapeutic relevance in the future, it would be very interesting to compare the effects on muscle activity and postural control of an unstable shoe with those of a therapeutic sensorimotor training. This would allow patients to receive optimal advice on which intervention is most promising for them.

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List of abbreviations

CoP	center of pressure
CoM	center of mass
CoG	center of gravity
ANOVA	analysis of variances
EMG	electromyography
NS	norm shoe
BF	barefoot
m.	musculus
mm.	musculi
MVC	maximum voluntary contraction
SL	single leg
DL	double leg
SD	standard deviation
BMI	body mass index
RMS	root mean square
η^2	partial eta squared
MDISTap	mean distance anterior-posterior
MDISTml	mean distance medio-lateral
RDISTap	root mean square distance anterior-posterior
RDISTml	root mean square distance medio-lateral
TOTEXap	total excursion anterior-posterior
TOTEXml	total excursion medio-lateral
MVELOap	mean velocity anterior-posterior

MVELOml	mean velocity medio-lateral
RANGEap	range anterior-posterior
RANGEml	range medio-lateral
GMAX	musculus gluteues maximus
GMED	musculus gluteus medius
BIFEM	musculs biceps femoris
GASMED	musculus gastrocnemius medialis
PERO	musculus peroneus longus
TIBA	musculus tibialis anterior
VASMED	musculus quadriceps femoris vastus medialis
TENSOR	musculus tensor fasciae latae

Appendix

A. Information sheet

Date: _______________

Name: ___________________________________ ID: ____

Age: ______

Sex: ☐ male ☐ female

Size: ____cm

Weight: ____kg

Dominant leg: ☐ left ☐ right

Shoe size: _____

Shoe weight: _____g

Leg axis:

-5	-4	-3	-2	-1	0	1	2	3	4	5

valgus varus

Foot arch:

-5	-4	-3	-2	-1	0	1	2	3	4	5

fallen arch high arch

Walking velocity: ____m/s

B. Listings

Gait analysis

```
clear; close; clc;

var_names_time = {'BL','FU'};
var_names_group = {'IG','KG'};
var_names_cond = {'BF','NS','X10D'};
fn_muscles = {'GMED', 'GMAX', 'BIFEM', 'GASMED', 'PERO', 'TIBA', 'VASM',
'TENSF'};

path_data_BL = {

'C:\Users\Bernhard\Desktop\X10D_coding\MasterarbeitDaten_Gang_aufbereitet
_mitOffsetKorrektur\BL_IG_BF\',...

'C:\Users\Bernhard\Desktop\X10D_coding\MasterarbeitDaten_Gang_aufbereitet
_mitOffsetKorrektur\BL_IG_NS\',...

'C:\Users\Bernhard\Desktop\X10D_coding\MasterarbeitDaten_Gang_aufbereitet
_mitOffsetKorrektur\BL_IG_X10D\',...

'C:\Users\Bernhard\Desktop\X10D_coding\MasterarbeitDaten_Gang_aufbereitet
_mitOffsetKorrektur\BL_KG_BF\',...

'C:\Users\Bernhard\Desktop\X10D_coding\MasterarbeitDaten_Gang_aufbereitet
_mitOffsetKorrektur\BL_KG_NS\',...

'C:\Users\Bernhard\Desktop\X10D_coding\MasterarbeitDaten_Gang_aufbereitet
_mitOffsetKorrektur\BL_KG_X10D\'};

path_data_FU = {

'C:\Users\Bernhard\Desktop\X10D_coding\MasterarbeitDaten_Gang_aufbereitet
_mitOffsetKorrektur\FU_IG_BF\',...

'C:\Users\Bernhard\Desktop\X10D_coding\MasterarbeitDaten_Gang_aufbereitet
_mitOffsetKorrektur\FU_IG_NS\',...

'C:\Users\Bernhard\Desktop\X10D_coding\MasterarbeitDaten_Gang_aufbereitet
_mitOffsetKorrektur\FU_IG_X10D\',...
```

```
'C:\Users\Bernhard\Desktop\X10D_coding\MasterarbeitDaten_Gang_aufbereitet
_mitOffsetKorrektur\FU_KG_BF\',...

'C:\Users\Bernhard\Desktop\X10D_coding\MasterarbeitDaten_Gang_aufbereitet
_mitOffsetKorrektur\FU_KG_NS\',...

'C:\Users\Bernhard\Desktop\X10D_coding\MasterarbeitDaten_Gang_aufbereitet
_mitOffsetKorrektur\FU_KG_X10D\'};

all_paths = [path_data_BL; path_data_FU];
    data = struct;
    for j = 1:2

        Counter_path = 1;

        for n = 1 : 2
            for o = 1 : 3

                file_tmp = all_paths(j,Counter_path);
                file = dir(file_tmp{1});

                for i = 3 : length(file)

                    work_file  = strcat(file_tmp,file (i).name);
                    xls_tmp = xlsread(work_file{1});


data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).GMED(i,
:) = xls_tmp(34:133,11);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).GMAX(i,
:) = xls_tmp(34:133,13);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).BIFEM(i
,:) = xls_tmp(34:133,15);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).GASMED(
i,:) = xls_tmp(34:133,17);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).PERO(i,
:) = xls_tmp(34:133,19);
```

```
data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).TIBA(i,
:) = xls_tmp(34:133,21);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).VASM(i,
:) = xls_tmp(34:133,23);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).TENSF(i
,:) = xls_tmp(34:133,25);


data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.StandphaseTO(i,1) = xls_tmp(2,2);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.Zweibeinstandphase(i,1) = xls_tmp(16:16,2);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.Fussrotation(i,1) = xls_tmp(19:19,2);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.Schrittlaenge(i,1) = xls_tmp(21:21,2);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.Schrittbreite(i,1) = xls_tmp(24:24,2);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.LaengeDoppelschritt(i,1) = xls_tmp(23:23,2);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.Geschwindigkeit(i,1) = xls_tmp(25:25,2);

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.Kadenze(i,1) = xls_tmp(31:31,2);
                    TO_tmp =
round(data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).P
arameter.StandphaseTO(i,1));


data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.peaksGMED(i,1) = max(xls_tmp(34:(34 + (TO_tmp)),12));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.peaksGMAX(i,1) = max(xls_tmp(34:(34 + (TO_tmp)),14));
```

```
data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.peaksBIFEM(i,1) = max(xls_tmp(34:(34 + (TO_tmp)),16));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.peaksGASMED(i,1) = max(xls_tmp(34:(34 + (TO_tmp)),18));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.peaksPERO(i,1) = max(xls_tmp(34:(34 + (TO_tmp)),20));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.peaksTIBA(i,1) = max(xls_tmp(34:(34 + (TO_tmp)),22));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.peaksVASM(i,1) = max(xls_tmp(34:(34 + (TO_tmp)),24));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.peaksTENSF(i,1) = max(xls_tmp(34:(34 + (TO_tmp)),26));


data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.meanGMED(i,1) = mean(xls_tmp(34:(34 + (TO_tmp)),12));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.meanGMAX(i,1) = mean(xls_tmp(34:(34 + (TO_tmp)),14));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.meanBIFEM(i,1) = mean(xls_tmp(34:(34 + (TO_tmp)),16));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.meanGASMED(i,1) = mean(xls_tmp(34:(34 + (TO_tmp)),18));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.meanPERO(i,1) = mean(xls_tmp(34:(34 + (TO_tmp)),20));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.meanTIBA(i,1) = mean(xls_tmp(34:(34 + (TO_tmp)),22));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.meanVASM(i,1) = mean(xls_tmp(34:(34 + (TO_tmp)),24));

data.(var_names_time{j}).(var_names_group{n}).(var_names_cond{o}).Paramet
er.meanTENSF(i,1) = mean(xls_tmp(34:(34 + (TO_tmp)),26));
```

```
            end
            Counter_path = Counter_path + 1;
        end
    end
end
```

Stance analysis

```
clear; close; clc;
var_names_time = {'BL','FU'};
var_names_group = {'IG','KG'};
var_names_ID = {'ID1', 'ID2', 'ID3', 'ID4', 'ID5', 'ID6', 'ID9', 'ID11',
'ID14', 'ID15', 'ID17', 'ID18', 'ID19', 'ID20', 'ID21', 'ID23', 'ID29',
'ID32', 'ID33', 'ID34', 'ID7', 'ID8', 'ID12', 'ID13', 'ID16', 'ID22',
'ID24', 'ID25', 'ID26','ID27', 'ID28', 'ID30', 'ID31'};
var_names_cond = {'BF_DL', 'BF_SL', 'NS_DL', 'NS_SL', 'X10D_DL',
'X10D_SL'};

data = struct();
path_data_BL = {
    'C:\Users\Bernhard\Desktop\X10D_coding\Stand\BL\IG\ID1\BF_DL\',...
    'C:\Users\Bernhard\Desktop\X10D_coding\Stand\BL\IG\ID1\BF_SL\',...
    'C:\Users\Bernhard\Desktop\X10D_coding\Stand\BL\IG\ID1\NS_DL\',...
    'C:\Users\Bernhard\Desktop\X10D_coding\Stand\BL\IG\ID1\NS_SL\',...
…};
path_data_FU = {
    'C:\Users\Bernhard\Desktop\X10D_coding\Stand\FU\IG\ID1\BF_DL\',...
    'C:\Users\Bernhard\Desktop\X10D_coding\Stand\FU\IG\ID1\BF_SL\',...
    'C:\Users\Bernhard\Desktop\X10D_coding\Stand\FU\IG\ID1\NS_DL\',...
    'C:\Users\Bernhard\Desktop\X10D_coding\Stand\FU\IG\ID1\NS_SL\',...
…};

all_paths = [path_data_BL; path_data_FU];
for i = 1:2
    Counter_path = 1;
    for k = 1 : 33
        for l = 1 : 6

            file_tmp = all_paths(i,Counter_path);
            file = dir(file_tmp{1});

            z=1;
            for m = 3 : 7
```

```
                work_file  = strcat(file_tmp,file (m).name);

                COPts = importfile(work_file{:,1},2500,17500);
                x = table2array(COPts(:,1));
                y = table2array(COPts(:,2));
                [Results_tmp(z,:), VariableName_tmp] =
STanalyzer(x,y,20,1000,'y');
                z = z + 1;
            end
            MeanResults_tmp_subj = mean(Results_tmp,1);
            data.(var_names_time{i}).(var_names_cond{l}).COPresults(k,:)
= MeanResults_tmp_subj;
            ID_pos = strfind(file_tmp{:,1},string('ID'));
            currentID = file_tmp{1,:}(ID_pos+2:ID_pos+3);
            data.(var_names_time{i}).(var_names_cond{l}).ID(k,:) =
currentID;

            Counter_path = Counter_path + 1;
        end
    end
```