Habilitation Thesis

Strength training: Towards subject specific modelling, individual internal loading conditions and design of exercises

Author(s):
Lorenzetti, Silvio

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Strength training: Towards subject specific modelling, individual internal loading conditions and design of exercises

presented by

Silvio Lorenzetti
Diplom Physiker UniBe
Dr. phil.-nat. UniBE
Dr. sc. ETH Zurich
born September 28 1974
citizen of Hallau, SH

Institut für Biomechanik
ETH Zurich

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1 Introduction

Strength training is used for injury prevention and rehabilitation, to enhance general fitness and to increase the level of performance in competitive sport (209). Besides these positive aspects, however, especially the lower limbs and the trunk are susceptible to injury during strength training (162). For the adaptation of the tissue, a certain level of functional loading is required to allow for biological adaptation. As a result, it is important to understand the internal loading conditions, as well as their modulators, in order to ensure for positive adaptation and avoid overload of the tissue.

1.1 Strength training

Strength or resistance training can include both machines and free weights (209). The aim in general is to have a high stress in the target muscle(s) over both eccentric and concentric muscle activity (107). There are several principles of load demands such as those presented by Stone et al. (209). The main idea is to load the muscle according to its capability and maintain the exposure during the workout as high as possible in order to start a positive adaptation process.

Already the Roman gladiators knew that strength training enhances muscle force, strength and power (Wiedemann). It is now known that strength training also supports the deposition of bone mass in adults (83), as well as reduces difficulties of performing daily tasks and enhances energy expenditure and body composition in older adults (104). In addition to these positive aspects for healthy individuals, strength training is known to be beneficial for people suffering neuromotor diseases or musculoskeletal deficits. As an example, strength training is beneficial following stroke (159), for patients with mild-to-moderate Parkinson’s disease (194), with osteoarthritis of the knee (197) and end-stage renal disease (89). Furthermore, strength training can be an effective countermeasure for some of the
adverse effects experienced by patients of many chronic diseases (105).

Recently, Buitrago et al. (23) studied the mechanical loading and physiological responses of the human body to four different resistance training methods. The authors concluded that physiological parameters such as oxygen consumption can be influenced with a change of loading patterns during exercising. Farinatti et al. (65) found that it might be beneficial for elderly, untrained women to first perform exercises for the larger muscle groups based on an analysis of the influence of changing repetition numbers and exercise order. Furthermore, Romero-Arenas et al. (187) analyzed the possibility of high-resistance circuit training to improve strength, cardiovascular parameters, bone mineral density and body composition in elderly people. It was shown that different loading patterns during circuit training can have positive training effects. Furthermore, Heggeîund et al. (91) found an advantage through maximizing loading for force development rate and maximal strength. Here, strength training intervention in aging population seems to lead to large improvements in maximal strength, walking time, and balance in both genders (101). All these aforementioned studies demonstrate the benefits of varying loading during strength training, and potential positive influences on the adaptation of the human body.

1.2 WHO recommendations

Based on the positive aspects of strength training, the World Health Organization (WHO) included strength training in their recommendation for health. In their white paper “Global recommendations on physical activity for health”, physical inactivity was identified as the fourth leading risk factor for global mortality (238). Based on Bauman et al. (11) and the Centers for Disease Control and Prevention CDC review (179) including 1598 papers, the WHO recommends that adults (18-64 years old) should engage in muscle-strengthening activities involving major muscle groups two or more
days a week to improve cardiorespiratory and muscular fitness, bone health and reduce the risk of non-communicable diseases (NCDs) (238), in addition to aerobic physical activity. The same recommendations for physical activity have been issued by the WHO for adults (65 years old and above) to maintain health and reduce the negative effects of aging.

1.3 Exposure & injury risk

Many people follow the recommendations of the WHO. In Switzerland, the motivation to engage in sports is 96.1% "health-related" according to a survey by Lamprecht et al. (124) (n=11000). Fitness training (14.0%) together with resistance training and bodybuilding (3.4%) are among the top five most popular sports for the Swiss population (124). Fourteen percent of the questioned sub-population had a contract with a fitness center, and an additional 7-8% were frequent visitors without a contract (124).

Despite the plethora of benefits for both healthy subjects and patients alike, one of the possible disadvantages of strength training, however, is musculoskeletal injury. According to a statistic analysis by the Bundesamt für Unfallforschung 2012 (Swiss governmental institute for accident research), 1.5% of all sports injuries are related to fitness training, gymnastics, or aerobics (222). Based on a questionnaire (n=6559), Müller (162) identified strength training as an injury risk for 2.5% for men and 1.6% for women with 8.1% of customers submitting complaints within a year. The main injury locations due to strength training were the shoulders (24.4%), the back (16.5%), the thighs (11%) and the knees (8.7%)(162). The authors identified leg press / squats with barbells (knees), the back extension (back), and bench press (shoulder) as potentially risky exercises. The main reason (68.4%) for injury was incorrect performance of the exercises, specifically overloading (45.6%) and incorrect execution (21.1%) (162). In order to practice strength training safely, it is important to perform the exercises
correctly (250). A key aspect of training is therefore to apply suitable loads
to the target structures without overloading or damaging them or the sur-
rounding soft tissues. However, relatively little is known about the internal
loading conditions, including both muscle and joint contact forces, that oc-
cur during most forms of strength training.

1.4 Exercise guidelines

Most of the existing guidelines on strength training are based on experience,
such as the exercise collection by Bill Pearl (178). The American College
of Sports Medicine (ACSM) published a position stand making recommenda-
dations for training in order to design a work-out that has the capability
to optimally achieve the aim of the training. However, the most recent po-
sition stand (as well as previous ones) has come under heavy criticism for
misrepresentation of research, lack of evidence and author bias (107). These
authors also published a list with suggestions for future research that in-
cluded the requirement for understanding load and different resistance, in
order to gain knowledge about a suitable intensity of effort (108).

For certain strength exercises, the National Strength and Conditioning
Association (NSCA) published a position paper including a review of the
literature (32). However, evidence-based recommendations on how to per-
form strength exercises are sparse. As an example, one of the guidelines by
the NSCA in order to reduce the injury risk of the knee is to keep the shin
as vertical as possible during squatting. This "rule" is only based on the
analysis of squatting in three individuals (6), and specifically on the data
from the most experienced lifter, who exhibited the lowest shear force at the
beginning of the exercise with his shin vertical.
1.5 Biomechanical studies

In order to provide sufficient evidence, studies about segment movement and loading during exercising, including the analysis of kinetics and kinematic parameters during strength training, are required to acquire sound evidence on the levels of loading that can be considers safe for each individuals musculoskeletal system and improve guidelines.

The study of human movement in general has been performed since the Antiquity. Historically, the analysis of human movement has been based on the derivation of kinematic and kinetic parameters and the development of new measurement techniques. Nigg and Herzog (168) summarized historical highlights in the development of "Biomechanics" starting in Antiquity (650 B.C. - 200 A.D.). In brief, Leonardo da Vinci (1452-1519) studied the parallelogram of forces and friction and analyzed the mechanics of human movement including joints, muscles, bones, ligaments, tendons and cartilage (168). Thereafter, the term "mechanics" was introduced as a subtitle of the book Two New Sciences by Galileo (1638). In 1680, Borelli (1608-1679), a colleague of Galileo, published a book called De Motu Animalium, analyzing the motion of animals in air and water. Two hundred years later, Muybridge used a battery of 12 automatic electro-photographic cameras to acquire images of humans and animals in motion (160). His studies, based on a series of photographs, were published in the books Animal Locomotion in 1887 and The Human Figure in Motion in 1901. In the famous book The biomechanics of sport techniques by James G. (87), biomechanics was defined as the science that is concerned with the forces that act on a human body and the effect these forces produce. In a more recent definition by Fung (73), biomechanics is considered as the study of mechanics applied to biology.

These definitions are still valid in modern-day research. Nowadays, inverse dynamics analysis is commonly used to calculate internal moments
and forces based on measured body motion and forces that act on the body. Whole body simulation software such as Opensim (Simbios project, National Institutes of Health) or Anybody (Anybody Technology A/S, Aalborg, DK) are available to derive the internal forces and moments based on measured data. The forces acting on the body can non-invasively be measured using force plates, position with Magnetic Resonance Imaging (MRI) and motion data can be acquired using optical motion capture. Unfortunately, the positions of skin markers for optical motion capture do not directly reflect the positions of the underlying bones, particularly for segments with more pronounced soft-tissue coverage (217). In contrast, MRI provides image data of internal structures without any soft-tissue artefacts or harmful x-ray exposure. However, the acquisition of image data during dynamic movement is limited using MRI. A new open MRI allows the imaging of, for example, the weight-bearing spine during a sitting posture with flexed and extended spine (57). This allows positions with a higher injury risk to be assessed, albeit statically. Furthermore, functional MRI (fMRI) is increasingly used in sports medicine and research (57). Ideally, a combination of fMRI and optical motion capture can be used to 1) determine the measurement error between the motion of the skin and the underlying bones in static positions, and 2) derive the motion of the bones during dynamic movements with real data, including soft-tissue artefacts. This provides high quality position data of the joints that is required for inverse dynamic calculations.

1.6 Conclusions & Aim

Although there is clear evidence that strength training has positive effects on activities of daily living, the cause-effect-chains from mechanical loading towards adaptation of the tissue, including physiological performance measures, is barely known. Therefore Beijersbergen et al. (14) suggested a paradigm shift from conventional outcome assessments to biomechanical
analyses including joint kinematics, kinetics and modelling.

For most strength exercises no evidence based instruction or guidelines are available, although people are performing strength training are eager to train correctly. A google search on the term "How to squat" provides 29'900'000 results mainly advice based on personal experience. There are also many very popular text books available such as "Keys to the inner universe" by Bill Pearl, that are purely based on practical knowledge. A number of more general rules exist that address the most appropriate kinematics and kinetics. A strength exercise is executed safely within the physiological range of motion (ROM) of a joint and by avoiding overloading of the human tissue (250). However, strength exercises should ideally be performed over the entire ROM to optimize the training effect (9).

Despite this fundamental requirement, clear evidence-based guidelines on how strength training can be performed with sufficient mechanical stimuli on the target areas without musculoskeletal overloading, are for many exercises lacking. Furthermore, it is important that there is sufficient mechanical stress on the target structure without overloading other parts of the body. Here, especially the influence of different types and execution forms of exercises on the resulting internal loading conditions is barely known.

Therefore this work aims to assess kinetic and kinematic data to calculate the loading conditions in the joint for the strength exercises squat, lunges, deadlifts and good mornings in order to provide evidence for guidelines, rational to choose suitable exercises, and insights in to the specific internal loading conditions on the joints, muscles and soft tissue structures.

1.7 Outline of the thesis

The present work starts with an analysis of the loading states of one of the primary injury sites, the knees and thighs. In Chapter 2, the influence of step length and angle of the frontal tibia on the loading of the leg was
studied during different types of split squat exercises. In order to analyse the influence of the "old" NSCA recommendation on the motion of the knee relative to the toes, the movement of the knee and hip, as well as the loading conditions during two different types of squat executions (Chapter 3) was assessed and the required measurement technique was developed. In chapter 4 this measurement technique was used to correlate biomechanical parameters during squat jumps, squats and imitation jumps of skijumpers with their performance in competition. Our findings regarding the squats showed that not only the knees but specially the movement and loading of the trunk to be of high importance. Since a suitable skin marker set was missing, we developed one to analyze the motion of the trunk during the two different types of squats (Chapter 5). However the accuracy of our new marker set remained open. For validation purposes of the newly developed marker set, different sitting positions were firstly analyzed using fMRI (Chapter 6), and in a second step, the skin marker set was attached during the fMRI measurements to analyze the associated soft-tissue artifacts (Chapter 7). The gained knowledge was then used to assess the motion and loading during two exercises for the trunk, namely good-mornings and deadlifts in chapter 8 focusing on the main body parts with a high injury risk. Finally, an overall conclusion, on-going work in the area, perspective and vision are presented.
The aim of this study was to quantify how step length and frontal tibia angle influence joint angles and loading conditions of the knee and the hip during the split squat exercise. It was found that the motion and loading conditions of the knee joint are influenced by changes in the frontal tibia angle and variations in step length. The results from the present work will allow coaches and therapists to adapt the split squat exercise to subject-specific knee motion and loading demands during training or rehabilitation.

A summary of this work was presented in a popular scientific sports magazine *Fit for Life*, 12, 2014.

**Contributions:** The data was gathered during the internship of PS. All authors contributed to the text of the manuscript. SL designed the study, supported and supervised the data evaluation and discussions about the evaluation and interpretation of the results.
Joint angles of the ankle, knee and hip and loading conditions during split squats

Pascal Schütz, Renate List, Roland Zemp, Florian Schellenberg, Silvio Lorenzetti

Institute for Biomechanics, ETH Zurich, Switzerland

Address correspondence to Silvio Lorenzetti, sl@ethz.ch

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2.1 Abstract

The aim of this study was to quantify how step length and the front tibia angle influence joint angles and loading conditions during the split squat exercise. Eleven subjects performed split squats with an additional load of 25% body weight applied using a barbell. Each subject’s movements were recorded using a motion capture system, and the ground reaction force was measured under each foot. The joint angles and loading conditions were calculated using a cluster-based kinematic approach and inverse dynamics modelling respectively. Increases in the tibia angle resulted in a smaller range of motion (ROM) of the front knee and a larger ROM of the rear knee and hip. The external flexion moment in the front knee/hip and the external extension moment in the rear hip, decreased as the tibia angle increased. The flexion moment in the rear knee increased as the tibia angle increased. The load distribution between the legs changed < 25% when split squat execution was varied. Our results describing the changes in joint angles and the resulting differences in the moments of the knee and hip will allow coaches and therapists to adapt the split squat exercise to the individual motion and load demands of athletes.

2.2 Introduction

A split squat, or a lunge with a barbell, is a multi-joint exercise of the lower body used to strengthen the gluteus maximus, iliopsoas, quadriceps, hamstrings, soleus, and gastrocnemius muscles (78). This closed chain kinetic exercise was developed to improve the function of the lower limbs (90). Walking forward lunges and jumping forward lunges have been shown to improve hamstring strength and sprint performance, respectively (111). However, split squats are more commonly used in rehabilitation settings, particularly after cruciate ligament reconstruction (64). Thus, it is important to understand how the mechanical demand is distributed across the
joints when evaluating the appropriateness of an exercise (70).

There are several types of lunge and split squat exercises. Some lunges are initiated with the feet in the same anterior position, followed by a step forward (forward lunge) (78) or a step backward (backward lunge) (115). Split squats are performed by anteriorly displacing one foot and maintaining a constant step length during repetitions (90). However, it appears prudent to perform split squats (62) since forward lunges (with step lengths of 1.01m and 0.84m) have shown to result in a significantly different peak ankle flexion angles but no significant differences in the peak flexion angles of the knee and hip (184). When examining forward lunges, it was found that trunk position is able to significantly affect the biomechanics of the lower extremities, (68) with different loading demands and joint angles thus providing a rationale for the selection of specific lunge variations (184). No differences in the angles of the knee and hip between men and women were observed during lunges (52). Additionally, no differences in the peak flexion angles for the ankle, knee, or hip of the anterior leg were observed with 0%, 12.5%, 25%, or 50% additional body weight (BW) load (185).

According to a fitness center accident report in Switzerland (162), most split squat-induced injuries occur due to customer error. A total of 21% of injuries were due to incorrect exercise execution, and 45.6% were due to overload (162), despite several available guidelines that describe the correct execution of split squats. The base of support for split squats is larger in the anterior/posterior direction than for the standard squat exercise thus allowing the subject to vary several performance parameters. Split squat guidelines focus particularly on step length, the weight distribution between the legs, and joint angles, such as the shank position. For step length, taking a moderately large forward step (115) or stepping forward as far as possible is recommended (178). A common guideline for achieving proper joint angles is to keep the front knee directly above the front foot (63).
However, during squatting, appropriate joint loading may require the knees to move slightly over the toes (72). It is expected that the subjects' weight should be distributed between the legs with either 50% on the front leg and 50% on the rear leg or 60% on the front leg and 40% on the rear leg. However, these guidelines are based on common experience rather than science-based evidence, and it remains unknown whether and under which movement execution the loading conditions generated from these exercises are appropriate for training versus rehabilitation.

In a study of different split squat execution methods, it was generally observed that the load on the knee of the front leg is large at high knee angles and smaller at low knee angles (63). Furthermore, the use of external weights by the subjects results in an increase in hip and ankle kinetics, but little change in the knee loading contribution (185). Recently, it was shown that the external joint moments of the knee and hip can be influenced by the step length and front knee position (203). In general, a strength exercise is executed safely within the physiological range of motion (ROM) of a joint and by avoiding overloading of the human tissue (250). However, strength exercises should ideally be performed over the entire ROM to optimize the training (9). For split squats it therefore follows that tibia angle and step length need to be chosen appropriately.

It was also reported that the knee joint forces and stress magnitudes at a given knee angle for the front leg are similar during the eccentric and concentric phases of the split squat (63). The same observation was made for the squat strength exercise (85; 140). The use of external weights by the subject results in an increase in hip and ankle kinetics and little change in the knee loading contribution (185). Although several split squat execution types and guidelines have been described, it is unclear how such factors as a change in step length or shank position influence the loading of the limbs. As a result, we hypothesized that the ROMs and the corresponding...
moments of the knees and hips are affected by step length and tibia angle. Therefore, the aim of this study was to compare the ankle, knee and hip ROM, the vertical load distribution between the front and rear legs, and the corresponding moments of the front and rear knee and hip during split squats of different step lengths and front tibia angles.

2.3 Methods

Subjects

Five female and six male students of human movement science were recruited for this study. The subjects had an average age of 24.9±2.5y, an average weight of 68.1±8.8kg, and an average ll of 90.3±4.5cm. The study was approved by the Eidgenössische Technische Hochschule Zurich Ethics Committee. Experience in weight training was mandatory for all subjects, but no specific sports activity background was required. The subjects were instructed to wear their normal sports shoes and to be nourished and hydrated.

Experimental Approach to the Problem

The subjects performed six repetitions of ten different split squat types with a standard additional load of 25% BW added using a barbell. Step lengths of 55%, 70%, and 85% of leg length (ll) as well as tibia angles of 60°, 75°, 90°, and 105° (tibia perpendicular to the ground) were evaluated (Figure 1). A larger tibia angle was found to correspond to increasing plantar flexion. All combinations of tibia angles and step lengths were performed, except at the tibia angle of 105°, where only a step length of 85% was executed. The hip-wide stance was calculated from the inter-ASIS distance (205). Each subject was instructed to keep the upper body erect and lunge as deeply as possible without touching the ground. The average time for one repetition was 2-3s, and the order of execution was randomized. To maintain the mass
distribution of the body, the applied additional load was a percentage of
the subject’s BW; however, this scheme resulted in different effort levels
across subjects. The hip-wide stances and step lengths were marked on the
ground. The ground reaction force of each foot and the motion of the body
were measured and analyzed. The dependent variables were the knee and
hip flexion angles and moments, and the independent variables were the step
length and tibia angle at the lowest point of the split squat.

Procedures

The subjects were instructed in the performance of split squats (Table 1).
Each subject was able to control tibia angles by checking a monitor that
displayed the position of the tibia with respect to reference lines (Figure
1). A video camera recorded a view of the sagittal plane from the left
side, which included the front leg and a plate with reference angles. This
setup was chosen to prevent restrictive forces from acting on the knee. The
kinematics of the split squat were evaluated using a 12-camera (MX40)
3D motion system (Vicon, Oxford, UK). Two Kistler force plates (type
9281B, 400x600mm, Winterthur, Switzerland), one for each foot, were used
to measure the ground-reaction forces, with each force plate specifically
calibrated using 56 points on a calibration grid (140) to allow accurate data
collection for use in inverse dynamics analyses. The kinetic and kinematic
data were recorded simultaneously at 2,000Hz and 100Hz, respectively. This
setup allowed for proper data collection of the inverse dynamics when each
foot was placed on a force plate. Each force plate was calibrated using
56 points on a calibration grid (140). The marker set consisted of 53 skin
markers for the legs and body (136; 242) and two for the barbells. Trained
personnel attached the markers with double-sided tape.

The center and axis of the ankle, knee, and hip joints were functionally
determined using standardized basic-motion tasks (136). The estimations
of joint rotations were based on a least-squares fit of redundant point clouds (74), a helical axis approach (243), and orthogonal, anatomically defined joint coordinate systems. Inverse dynamic calculations based on the position of the body and the ground reaction forces to calculate the moments. The weights of the thigh, shank, and foot were assumed to be 12.64%, 4.24%, and 1.32% of the BW for males, respectively, and 13.81%, 4.88%, and 1.27% of the BW for females, respectively (51). The center of gravity was assumed to be on the line between the joint centers within 40.65% and 41.04% of the thigh and shank lengths, respectively, for males and within 38.81% and 40.98%, respectively, for females (51). Joint moments were then calculated and normalized to the BWs of individual subjects. The calculations of the mean maximum moments and angles were based on the maximum values for each subject. The ROM was defined as the difference between the maximal flexion (plantar flexion for the ankle) and minimal flexion (maximal extension; plantar extension for the ankle) sagittal joint angle during a repetition. The vertical load distribution between the front and rear legs was calculated by dividing the vertical force of the front leg by the sum of the vertical forces of both legs at the instant of the maximal moment of the front knee. All calculations were performed in MATLAB version R2010 (Natick, Massachusetts, USA). Only the sagittal moments are presented because the moments in the frontal and transverse planes were less reproducible (112). The knee and hip moments were defined as positive for flexion. The measurements were randomly distributed relative to the time of day and year.

Statistical analyses

Six valid split squat repetitions for each execution type were averaged for statistical calculations. A linear mixed model was used to evaluate how the ROM of the ankle, knee, and hip joints; the maximal moments of the knee and hip of each leg; and the vertical load distribution between the front and
Figure 1: The measurement setup for split squats included 1) an angle meter and 2) a screen that projected a side view of the subject captured using 3) a video camera. This surveillance allowed the subject to control the angle of the front tibia. 4) Individual marks designating the step length were placed on the force plates.

Table 1: Standardized instructions for split squat performance.

<table>
<thead>
<tr>
<th>Instructions</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Put the barbell on the trapezius muscle, and hold it comfortably in your hand.</td>
</tr>
<tr>
<td>2. Stand upright with your feet on the hip-wide position marks. Step forward with the right leg to the specified marked step length.</td>
</tr>
<tr>
<td>3. Start with both knees in full extension.</td>
</tr>
<tr>
<td>4. Lunge as deeply as possible without touching the ground with the rear leg.</td>
</tr>
<tr>
<td>5. The specified tibia angle should be achieved at the deepest position of the split squat.</td>
</tr>
<tr>
<td>6. Keep the upper body erect during the exercise.</td>
</tr>
<tr>
<td>7. Hold tension in the core muscles during the execution of the split squat.</td>
</tr>
<tr>
<td>8. Perform the downward and upward movements of the split squat at the same normal speed.</td>
</tr>
</tbody>
</table>
rear legs differed with execution type. The model used tibia angles, step lengths, and interaction of step length and tibia angle as fixed effects and subjects as random effects. When the tibia angle and step length interaction was not significant for a specific ROM, moment or vertical load distribution, Bonferroni corrected pairwise comparisons were performed, except in cases where the interaction between the two parameters, tibia angle and step length was significant. The significance level was set at $P < 0.05$. The statistical evaluation of the data was performed using IBM SPSS software version 19 (SPSS AG, Zurich, Switzerland).

2.4 Results

**Kinematics** In general, the largest ROM was observed in the knee joints of the front and rear legs (Table 2). The maximal ROM of the front leg was observed for a step length of 85% ll and a tibia angle of 60°, which resulted in the lowest observed ROM in the rear knee. Increasing the tibia angle decreased the ROM of the front knee and increased the ROM of the rear knee. The ROMs of the front and rear knee varied for different execution types by a factor of nearly two. While the ROM of the front ankle decreased as the angle of the front tibia increased, no difference in the ROM of the rear ankle joint was observed (Table 2). The ROM of the hip joint of the front leg was larger than that of the rear leg (Figure 2 and 3). While the joint angle of the front hip always in the flexion and was dependent on the step length and tibia angle, the hip angle of the rear leg changed in both flexion and extension (Figure 2). Split squats with a step length of 55% ll and a tibia angle of 90°resulted in the largest hip ROM in both legs (Figure 2). The interactions of the tibia angles and the step lengths were significant for the ROM of the hip of both the front and rear legs (Table 3).

**Kinematics** The vertical load distribution between the front and rear legs was different for different step lengths and tibia angles (Figure 3 and
Table 2: Range of motion (ROM [°], positive for flexion) for the ankle, knee, and hip joints of the front and rear legs during split squats with different step lengths. Split squat step lengths of $l_1=55\%$ leg length (ll), $l_2=70\%$ ll, and $l_3=85\%$ ll and four tibia angles (a) were evaluated. Significant differences in the ROMS for different step lengths and tibia angles are marked with a star. For the ROMs of the hip of the front and rear legs, there were significant interactions of the tibia angles and step lengths. Therefore, a Bonferroni corrected pairwise comparison could not be performed.

<table>
<thead>
<tr>
<th></th>
<th>a 60°</th>
<th>a 75°</th>
<th>a 90°</th>
<th>a 105°</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>ankle front leg</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>l1</td>
<td>$41.2 \pm 3.4$</td>
<td>$32.8 \pm 3.1$</td>
<td>$22.8 \pm 4.4$</td>
<td></td>
</tr>
<tr>
<td>l2</td>
<td>$43.4 \pm 3.9$</td>
<td>$34.1 \pm 3.1$</td>
<td>$22.9 \pm 3.7$</td>
<td></td>
</tr>
<tr>
<td>l3</td>
<td>$45.1 \pm 2.8$</td>
<td>$36.4 \pm 2.2$</td>
<td>$25.1 \pm 3.1$</td>
<td>$13.2 \pm 4.3$</td>
</tr>
<tr>
<td><strong>ankle rear leg</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>l1</td>
<td>$11.6 \pm 6.7$</td>
<td>$10.9 \pm 9.4$</td>
<td>$12.4 \pm 7.9$</td>
<td></td>
</tr>
<tr>
<td>l2</td>
<td>$11.6 \pm 8.2$</td>
<td>$12.2 \pm 9.6$</td>
<td>$13.5 \pm 9.2$</td>
<td></td>
</tr>
<tr>
<td>l3</td>
<td>$14.8 \pm 6.7$</td>
<td>$18.8 \pm 9.6$</td>
<td>$20.7 \pm 10.3$</td>
<td>$19.4 \pm 9.2$</td>
</tr>
<tr>
<td><strong>knee front leg</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>l1</td>
<td>$91.1 \pm 6.9$</td>
<td>$82.6 \pm 7.8$</td>
<td>$71.8 \pm 7.9$</td>
<td></td>
</tr>
<tr>
<td>l2</td>
<td>$91.9 \pm 6.6$</td>
<td>$81.8 \pm 6.5$</td>
<td>$70.4 \pm 7.8$</td>
<td></td>
</tr>
<tr>
<td>l3</td>
<td>$95.7 \pm 8.8$</td>
<td>$83.4 \pm 7.3$</td>
<td>$71.9 \pm 8.2$</td>
<td>$53.4 \pm 7.6$</td>
</tr>
<tr>
<td><strong>knee rear leg</strong></td>
<td></td>
<td></td>
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<td></td>
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<tr>
<td>l1</td>
<td>$82.2 \pm 7.9$</td>
<td>$85.9 \pm 13.9$</td>
<td>$89.2 \pm 15.4$</td>
<td></td>
</tr>
<tr>
<td>l2</td>
<td>$64.3 \pm 6.8$</td>
<td>$69.8 \pm 11.6$</td>
<td>$80.3 \pm 11.7$</td>
<td></td>
</tr>
<tr>
<td>l3</td>
<td>$43.6 \pm 10.8$</td>
<td>$50.1 \pm 11.9$</td>
<td>$62.6 \pm 7.0$</td>
<td>$66.6 \pm 9.5$</td>
</tr>
<tr>
<td><strong>hip front leg</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>l1</td>
<td>$42.8 \pm 8.7$</td>
<td>$46.0 \pm 10.3$</td>
<td>$50.5 \pm 9.5$</td>
<td></td>
</tr>
<tr>
<td>l2</td>
<td>$40.4 \pm 5.8$</td>
<td>$41.1 \pm 7.0$</td>
<td>$46.5 \pm 9.3$</td>
<td></td>
</tr>
<tr>
<td>l3</td>
<td>$42.8 \pm 8.1$</td>
<td>$39.2 \pm 7.5$</td>
<td>$39.4 \pm 7.6$</td>
<td>$37.5 \pm 6.4$</td>
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<td>$19.7 \pm 6.6$</td>
<td>$26.3 \pm 9.3$</td>
<td>$40.1 \pm 9.5$</td>
<td></td>
</tr>
<tr>
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<td>$14.1 \pm 6.1$</td>
<td>$29.1 \pm 7.6$</td>
<td></td>
</tr>
<tr>
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<td>$10.7 \pm 2.7$</td>
<td>$7.0 \pm 2.8$</td>
<td>$10.2 \pm 4.1$</td>
<td>$23.1 \pm 7.4$</td>
</tr>
</tbody>
</table>
Table 3: P-values of the linear mixed model of the two parameters step length and tibia angles as well as their interactions. The constant term of the linear mixed model was highly significant (Pvalue > 0.001) for all ROMs and moments.

<table>
<thead>
<tr>
<th>ROM</th>
<th>step length</th>
<th>tibia angle</th>
<th>interaction</th>
</tr>
</thead>
<tbody>
<tr>
<td>ankle</td>
<td>&lt;0.001*</td>
<td>&lt;0.001*</td>
<td>0.63</td>
</tr>
<tr>
<td>ankle</td>
<td>&lt;0.001*</td>
<td>&gt;0.102</td>
<td>0.398</td>
</tr>
<tr>
<td>knee</td>
<td>0.221</td>
<td>&lt;0.001*</td>
<td>0.746</td>
</tr>
<tr>
<td>knee</td>
<td>&lt;0.001*</td>
<td>&lt;0.001*</td>
<td>0.123</td>
</tr>
<tr>
<td>hip</td>
<td>&lt;0.001*</td>
<td>0.023*</td>
<td>0.020*</td>
</tr>
<tr>
<td>hip</td>
<td>&lt;0.001*</td>
<td>&lt;0.001*</td>
<td>&lt;0.001*</td>
</tr>
<tr>
<td>Moment</td>
<td>knee</td>
<td>&lt;0.001*</td>
<td>0.254</td>
</tr>
<tr>
<td>knee</td>
<td>0.038*</td>
<td>&lt;0.001*</td>
<td>0.027*</td>
</tr>
<tr>
<td>hip</td>
<td>&lt;0.001*</td>
<td>&lt;0.001*</td>
<td>0.361</td>
</tr>
<tr>
<td>hip</td>
<td>&lt;0.001*</td>
<td>&lt;0.001*</td>
<td>0.005*</td>
</tr>
</tbody>
</table>

Table 4: Vertical weight distribution between the front and rear legs. Split squat step lengths of l1=55% of the leg length (ll), l2=70% ll, and l3=85% ll and four tibia angles (a) were evaluated. Significant differences in the vertical weight distribution between the legs for different step lengths and tibia angles are marked with a star.

<table>
<thead>
<tr>
<th>a 60°</th>
<th>a 75°</th>
<th>a 90°</th>
<th>a 105°</th>
</tr>
</thead>
<tbody>
<tr>
<td>l1</td>
<td>0.66 ± 0.03</td>
<td>0.63 ± 0.03</td>
<td>0.57 ± 0.04</td>
</tr>
<tr>
<td>l2</td>
<td>0.69 ± 0.02</td>
<td>0.64 ± 0.02</td>
<td>0.60 ± 0.03</td>
</tr>
<tr>
<td>l3</td>
<td>0.70 ± 0.04</td>
<td>0.64 ± 0.03</td>
<td>0.60 ± 0.03</td>
</tr>
</tbody>
</table>
Table 5: Maximal normalized external moments ([Nm/kg], positive for external flexion) of the knee and hip joints during split squats with different step lengths. Split squat step lengths of $l_1=55\%$ leg length (ll), $l_2=70\%$ ll, and $l_3=85\%$ ll and four tibia angles (a) were evaluated. Significant differences in the ROMS for different step lengths and tibia angles are marked with a star. For the ROMs of the hip of the front and rear legs, there were significant interactions of the tibia angles and step lengths. Therefore, a Bonferroni corrected pairwise comparison could not be performed.

<table>
<thead>
<tr>
<th></th>
<th>a 60°</th>
<th>a 75°</th>
<th>a 90°</th>
<th>a 105°</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>knee front leg</strong></td>
<td></td>
<td></td>
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<td></td>
<td>![star]</td>
<td>![star]</td>
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</tr>
<tr>
<td>11</td>
<td>1.46 ± 0.14</td>
<td>1.38 ± 0.17</td>
<td>1.19 ± 0.22</td>
<td></td>
</tr>
<tr>
<td>12</td>
<td>1.40 ± 0.17</td>
<td>1.33 ± 0.20</td>
<td>1.14 ± 0.19</td>
<td></td>
</tr>
<tr>
<td>13</td>
<td>1.28 ± 0.12</td>
<td>1.19 ± 0.17</td>
<td>1.13 ± 0.21</td>
<td>0.80 ± 0.16</td>
</tr>
<tr>
<td><strong>knee rear leg</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>11</td>
<td>1.57 ± 0.15</td>
<td>1.68 ± 0.19</td>
<td>1.73 ± 0.22</td>
<td></td>
</tr>
<tr>
<td>12</td>
<td>1.45 ± 0.14</td>
<td>1.61 ± 0.17</td>
<td>1.71 ± 0.21</td>
<td></td>
</tr>
<tr>
<td>13</td>
<td>1.41 ± 0.17</td>
<td>1.57 ± 0.18</td>
<td>1.81 ± 0.20</td>
<td>1.82 ± 0.24</td>
</tr>
<tr>
<td><strong>hip front leg</strong></td>
<td></td>
<td></td>
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</tr>
<tr>
<td>11</td>
<td>1.72 ± 0.30</td>
<td>1.56 ± 0.30</td>
<td>1.42 ± 0.30</td>
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<tr>
<td>12</td>
<td>1.92 ± 0.20</td>
<td>1.69 ± 0.14</td>
<td>1.71 ± 0.32</td>
<td></td>
</tr>
<tr>
<td>13</td>
<td>2.27 ± 0.38</td>
<td>1.96 ± 0.31</td>
<td>1.82 ± 0.28</td>
<td>1.72 ± 0.29</td>
</tr>
<tr>
<td><strong>hip rear leg</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>11</td>
<td>-1.14 ± 0.23</td>
<td>-0.91 ± 0.19</td>
<td>-0.54 ± 0.17</td>
<td></td>
</tr>
<tr>
<td>12</td>
<td>-1.45 ± 0.30</td>
<td>-1.24 ± 0.25</td>
<td>-0.91 ± 0.23</td>
<td></td>
</tr>
<tr>
<td>13</td>
<td>-1.77 ± 0.23</td>
<td>-1.69 ± 0.14</td>
<td>-1.49 ± 0.31</td>
<td>-1.11 ± 0.27</td>
</tr>
</tbody>
</table>
Table 4). The highest observed value occurred with a step length of 85% ll and a tibia angle of 60°, which was 25% larger than the smallest observed value that occurred with the same step length and a tibia angle of 105°.

In all conditions, the largest moments of the knee occurred at the deepest position of the body with the largest knee flexion angles (Figure 2). The maximal observed joint moments occurred in the hip of the front leg with a tibia angle of 60° and a step length of 85% ll (Table 5). As the tibia angle increased, the moment decreased in the front knee and hip. The highest moments of the rear and front hips were observed with the largest step length and smallest tibia angle, whereas the smallest moment occurred at a tibia angle of 90° and a step length of 55% ll. Overall, the maximal and minimal moments were different for the knee and hip joints of both legs (Table 5). Bonferroni corrected pairwise comparisons indicated that the moments of the knee and hip joints of the front leg were influenced by the type of split squat execution (Table 5). The interactions of the tibia angles and step lengths were significant for the moments of the knee and hip joint of the rear leg (Table 3). In the rear hip, the moment was influenced by more than a factor of three. The execution type had the least effect on the rear knee, for which the influence on the moments was only 22%. No differences in the front tibia angles were observed for different split squat step lengths.

2.5 Discussion

The biomechanical characteristics of many exercises used for rehabilitation and strength training in sports medicine require investigation (185).

In practice, several guidelines for the performance of a proper split squat are in use, and focus mainly on the step length and weight distribution between the two feet. To the best of the authors’ knowledge, this study is the first to focus on the ROM of the knees, and hips and corresponding joint moments during split squats. We hypothesized that the ROM of the ankles,
knees, and hips and the moments of the knee and hip are affected by split squat execution. Our study found that the ROM of the ankle of the rear leg remained constant and that the ROM and moments of all other joints were influenced by the type of split squat. In general, the observed ROM of the knee is in agreement to the knee flexion angle of 70-90 ° with a step length of approximately 84 cm reported in the study of Escamilla and co-workers (62). While no differences in the knee angles were observed at two different step lengths in the study of Riemann and co-workers (184), in the present study, the ROM of the frontal knee was significantly different with the step length and even more with different tibia angles. Bonferroni corrected pairwise comparisons showed that the ROM of the ankles and knees were influenced by the type of split squat execution. Based on our experimental set-up, in which the subjects controlled the angle of the front tibia, it is not surprising that the ROM of the front ankle was affected by the type of split squat. At
Figure 3: Relation between average tibia angle of the front leg [°] and knee moment [Nm/kg] of the rear leg (top) as well as the front leg (bottom) for each step length and front tibia angle.
a given step length, the angle of the front tibia at the deepest position of the split squat was dependent upon the coordination of the joint angles of the ankle, knee, and hip joints. Although the vertical ground reaction force was larger in the front knee compared to the rear knee, the moment produced in the front knee was smaller due to a smaller external lever arm. Escamilla and co-workers (63) concluded that to reduce the patello-femoral force and stress magnitude between 0° and 90°, it may be prudent to perform a split squat with a long step length. Based on our data, their recommendations are appropriate for the front knee but result in a higher moment in the front hip. However, the highest observed moment in our study in the front hip is still lower than the moment during running (15), which is an indication that this loading of the front hip is safe for healthy subjects.

Based on the kinetic and kinematic findings, no evidence was found to support the common guideline that the knee should not move over the toes. This is in agreement with the findings during squatting (72; 140). Of course, increased anterior lean of the front knee due to a smaller tibia angle resulted in a larger flexion moment in the front knee. However, the calculated flexion moments in the rear knee are larger compared to the front knee in all conditions, and are decreasing with a more anterior path of the front knee. Therefore, greater care should be taken on the rear leg compared to the front one. With respect to the flexion moment of the knee, the current guidelines are therefore only valid for the front limb, but induce a larger loading on the rear knee. In one study, split squats performed by 20 elderly subjects with no additional weight, resulted in a peak hip moment of 1.25±0.34Nm/kg during descent and 1.31±0.36Nm/kg during ascent (70). In the same study, maximal peak moments of 0.84±0.23Nm/kg during descent and 0.81±0.22Nm/kg during ascent were observed in the knee. The hip moments measured in our study were at least 10% larger than those reported in the previous study of Flanagan and co-workers (70),
whereas the knee moments of the elderly subjects were at the lower limit of those measured in our study. However, the elderly subjects performing split squats in the study of Flanagan et al. (70) were able to select the step length (which was performed “as far as comfortable”) and speed, and were allowed to hold a handle to maintain their balance. A limitation of our study was the chosen 25% BW extra load on the barbell, since it was shown that the kinetics during split squat of the ankle, hip, and, to a lesser extent, the knee were influenced by the additional load of 0 to 50% BW (185). In contrast, the kinematics of these joints were not influenced. Thus, for future studies, it can be hypothesized that a change in additional load would change the moments of the joints, whereas the ROM would remain similar. The chosen extra weight in this study was, for most exercises, below the load used for general strength training in healthy subjects and above the weight chosen in aerobic strength training where up to 100 repetitions are performed. As a result, the specific conditions assessed in this study could be considered appropriate for rehabilitation. Split squats are a more complex exercise than standard squats because the legs do not work in parallel; the joint moments of the rear and front legs can occur in opposite directions. Thus, during a repetition, the hip flexors of the rear leg and the hip extensors of the front leg must work simultaneously. Another difference is that the moment of the front knee changes direction during split squats. When the subject is in the upper squat position, the moment of the front knee is different from that in the squat. The moments are similar when the subject is in the lower squat position. Split squats can result in higher loading of the knee joint and a decrease in spinal loading relative to squats due to a smaller additional barbell load.16 For the knee and hip of the front leg, the loading during split squats is similar to that of squats, whereas the loading of the rear leg is very different between the two exercises. There are other ways to adapt the loading and modify the angles of the knees and hips for split squats.
than for squats. This variety of potential approaches allows coaches and therapists to customize the exercise for a specific athlete or patient. In this study, the ROM and moments of the ankle, knee, and hip were determined for split squats with three different step lengths and four different front tibia angles. At the same step length, the load distribution between the two legs can change by up to 25% for different front tibia angles. The ROM during different split squats varied by a factor of 1.5 to 5, whereas the moments differed by a factor of only 1.2 to 3. These results illustrate the importance of choosing the right split squat step length and front tibia angle. In order to execute the exercises safely, coaches and therapists need to adapt the loading and the ROM to the subjects in order to avoid injury. A tibia angle of 60° should be chosen for high loading of the front leg. Please note that during this specific form of execution, the front knee goes beyond the toes. A larger tibia angle reduces the moments of the front knee and the front hip and seems to enhance the moment of the rear knee. A step length of 55% ll maximizes the moment of the front knee, whereas a step length of 85% ll maximizes the front hip moment. Performing split squats with a larger tibia angle leads to a smaller ROM of the front knee and larger ROM of the rear knee. A step length of 85% ll results in a smaller ROM of the rear knee compared with a step length of 55% ll. A large tibia angle and a large step length is preferred to minimize the loading and ROM of the front knee. Higher moments and a larger ROM make split squats more demanding than squats. It is important to note that the loading of the rear leg cannot be neglected during split squats, but since the additional barbell loading is generally smaller, spinal loading is lower. Furthermore, moment-angle plots allow for the quantification of movement and loading during the performance of split squats, providing quantified information for coaches and therapists to select the proper execution type to meet the needs of the athlete or patient, thus facilitating the design of efficient training programs.
that do not overload the body.

Acknowledgments

The technical support from Hans Gerber and Marco Hitz during the measurements of this study is gratefully acknowledged.
In this paper, the loading states of the knee and the hip joints were studied during different executions, namely, restricted and unrestricted squat strength exercises. The inverse dynamics method was adopted to determine the joint forces and moments in the knee and the hip joints based on input data from force plates. Because inverse dynamics is sensitive to uncertainties in the input data, a method is presented in this paper to minimize the error of the center or pressure measured by the force plate. The unrestricted squat was found to result in higher loads and Range of Motion (ROM) of the knee joint and smaller loads on the back compared to restricted squats. The results of the present biomechanical analysis suggest that the unrestricted squat should be recommended during strength training for maximal training effects with minimal injury risks.

This paper is in the top 25% of the total research output scored by Altmetric (Wolters Kluwer). The work was funded by the Bundesamt für Sport (BASPO).

The findings of the present work related to squatting were published in a popular non-scientific journal to promote the results amongst strength training practitioners (Lorenzetti, S. 2013, Bewegung und Belastung bei der Kniebeuge, *Fitness Tribune*, 141, 54-55.

**Contributions:** Within this project, Turgut Gülay and Mirjam Stoop performed their master thesis. SL designed the study, applied for funding, helped with and supervised the data acquisition, evaluation and interpretation. All authors provided text for the final manuscript. Alex Stacoff was in an investigator in an early stage of this project.
Comparison of the angles and corresponding moments in the knee and hip during restricted & unrestricted squats

Silvio Lorenzetti, Turgut Gülay, Mirjam Stoop, Renate List, Hans Gerber, Florian Schellenberg and Edgar Stüssi

Institute for Biomechanics, ETH Zurich, Switzerland

Address correspondence to Silvio Lorenzetti, sl@ethz.ch

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3.1 Abstract

The aim of this study was to compare the angles and corresponding moments in the knee and hip during squats. Twenty subjects performed restricted and unrestricted squats with barbell loads that were 0, 1/4, and 1/2 their body weight. The experimental setup consisted of a motion capture system and 2 force plates. The moments were calculated using inverse dynamics. During the unrestricted squats, the maximum moments in the knee were significantly higher, and those in the hip were significantly lower than during restricted squats. At the lowest position, the maximum knee flexion angles were approximately \(96^\circ\) for the restricted and approximately \(106^\circ\) for the unrestricted techniques, whereas the maximum hip flexion angle was between \(95^\circ\) and \(100^\circ\). The higher moments in the hip during restricted squats suggest a higher load of the lower back. Athletes who aim to strengthen their quadriceps should consider unrestricted squats because of the larger knee load and smaller back load.

3.2 Introduction

The squat is one of the classic exercises performed in strength training to improve the athlete’s ability to forcefully extend the knee and hips, and it can enhance performance in many sports (32). It is a common component of lower extremity rehabilitation and a closed kinematic chain exercise (206). Furthermore, the squat is one of the three disciplines used in powerlifting. Because of weight overload or improper execution, squats are associated with one of the highest risks of injury during strength training at Swiss fitness centers (162). Fifty percent of all injuries during fitness training involve either the lower extremities or the back. Lower back problems are an important issue in modern society (4). Therefore, the proper execution of the squat is important to avoid injury and to maximize the positive effects of training (27; 32; 49; 72). Low-to-moderate posterior shear forces are gener-
ated throughout the squat, which are restrained primarily by the posterior cruciate ligament (58). An analysis of 3 individuals found that the greatest magnitude of shear force occurred at the beginning of the exercise, when the subject initially bends the knee (subject 1: 104.5 kg, subject 2: 152.8 kg, subject 3: 161.2 kg), whereas the smallest posterior shear forces were reported at the lowest point (subject 1: 60.8 kg, subject 2: 12.6 kg, subject 3: 114.0 kg) (6). On the basis of these findings, the author associated the anterior translation of the knees with the greatest posterior shear force (subject 3). The lowest shear force, which was observed in subject 2, was attributed to the strength and experience of this lifter (6). The author concluded that anterior translation of the knee beyond the toes creates shear forces that might injure the knee. The highest calculated shear force in the knee during a squat with free weights and a squat machine was 1,350 N (5). However, this shear force is below the maximum tolerated load for healthy cruciate ligaments, which has been determined to be between 2160 ± 157 N (244) and 4000 N for young active people (181). A second argument favoring knee position restriction during squats is that the pressure between the patella and the femur increases with knee flexion (59; 169; 170). In general, this pressure is within the tolerable load limit (245). A cadaver study of 12 loaded knees showed a decrease of patellofemoral contact pressure and force by a factor of 2 at a knee flexion angle of 120° compared with a knee flexion angle of 90° (103). When studying the influence of the speed of squat execution, it was reported that when using a constant speed for descents and varying speed for ascents, there was a trend toward higher knee sagittal moments at higher speeds (88). According to Newton’s second law (166), the load conditions are expected to be higher during the acceleration of the body and extra load. However, the experimental data for repetition period of 1.8 seconds showed only small differences in the load condition (within the range of a few percent) during squat acceleration and deceleration (85).
The differences reported for restricted and unrestricted squats are based on the knee flexion angle and moments at the lowest point; however, at the lowest position of the squat, the knee flexion angle depends largely on the technique and the unique anatomy of the subject. One study (72) observed that the maximum moment for the knee decreases by a factor of 0.78 when subjects perform restricted squats, whereas the maximum moment in the hip increases by a factor of 10.7. The reported shear forces were smaller at the lowest position than during the rest of the repetition. A common guideline for the barbell squat is to prevent the knees from moving out beyond a vertical line over the toes. The instructions for proper knee position during squats, which were established in Europe by the National Strength and Conditioning Association (NSCA) (32; 72), are based on previous studies (6; 154; 153). The NSCA position paper (32) further states that, with some exceptions, the shin should remain as vertical as possible. Hence, squat kinematics and kinetics form the basis for the recommendations regarding the proper execution of the exercise (199). As far as we know, no literature is available presenting the sagittal knee and hip angles and the corresponding moments during the entire squat movement. From a practical point of view, this information is required to judge whether restriction of the knee during squatting is an important guideline. Therefore, the load condition on the knees and hips should be investigated not only at the lowest position but also at equal knee flexion angles throughout the squat movement. This study aimed to compare the knee and hip flexion angles and their corresponding moments throughout the entire repetition, especially at a knee flexion angle of 60° and at the lowest position during unrestricted and restricted squats. We hypothesized that the hip flexion angle and the moments in the knee and hip differ between restricted and unrestricted squats at an angle of 60° and at the highest flexion angle for the knee.
3.3 Methods

Experimental Approach to the Problem

The subjects performed restricted and unrestricted squats with loads of 0, $\frac{1}{4}$, and $\frac{1}{2}$ their body weight (BW) using a barbell, according to the instructions in Table 1. To maintain the mass distribution of the body and the extra load, the extra load was chosen as a percentage of the subject’s BW. However, this results in a different effort level across subjects. For the restricted squats, the subject was able to control the position of the knee with respect to the visible vertical line in front of their toes. This setup was chosen to avoid forces acting on the knee from the restriction. The ground reaction force of each foot and the motion of the body were measured and analyzed. The dependent variables were the knee and hip flexion angles and moments, and the independent variables were the execution type and the extra load.

Subjects

Ten female and 10 male subjects, all students of human movement science, with an average age of 24 ± 4 years, an average weight of 66 ± 12 kg, and an average height of 1.73 ± 0.08 m were recruited for this study. The study
was approved by the Eidgenössische Technische Hochschule Zurich Ethics Committee. All subjects signed informed consent documents. Experience in weight training was required for all subjects. The subjects had no specific sports activity background. The subjects were instructed to wear their normal sport shoes and to have a reasonable nutrition and hydration level.

**Procedures**

For the restricted squat, the knee was not allowed to go beyond a vertical plane over the toes. To control the motion of the knee for the restricted squat, a video camera view of the sagittal plane from the left side, which includes the knees and toes, was projected onto the wall in front of the subject (Figure 1). On the right side of the subject, a visible vertical line was placed at the toes. The kinematics of the squat were evaluated using a 12-camera (MX40) 3D motion system (Vicon, Oxford, United Kingdom). Two force plates (type 9281B, 400 x 600 mm; Kistler, Winterthur, Switzerland), one for each foot, were used to measure the ground reaction forces. The kinetic and kinematic data were recorded simultaneously at 2000 and 50 Hz, respectively. This setup allows for proper data collection for the inverse dynamics when each foot is placed on one force plate. Each force plate was calibrated using 56 points on a calibration grid (47). A weight of 18 kg was applied to each of these points. In addition, centerpunched aluminum plates (4 cm²) were fixed with doubled-sided tape at each point on the calibration grid. Using a rod with a metal tip, forces were applied over several angles (with a maximum of 30° from the vertical axis) to the center of the aluminum plates. This procedure resulted in a correction matrix for the center of pressure (COP) of the 56 points. Based on a fourth order polynomial-fit interpolation, each point of the force plate was corrected. The mean errors for the calibration grid points ranged from 0.8 to 19.8 mm and were eventually reduced to a range of 0.04 to 2.2 mm (47). The marker set
consisted of 53 skin markers for the legs and the body (7; 134; 242), 22 for the back (135), and 2 for the barbell. Trained personnel fixed the markers with double-sided tape.

The center and axis of the ankle, knee, and hip joints were functionally determined using standardized basic motion tasks (7; 135). The estimations of joint rotations were based on a least squares fit of redundant point clouds (74), a helical axis approach (243), and orthogonal anatomically defined joint coordinate systems. To calculate the moments, we performed inverse dynamic calculations based on the position of the body and the ground reaction forces. The weights of the thigh, shank, and foot were assumed to be 12.64, 4.24, and 1.32% of the BW for men and 13.81, 4.88, and 1.27 for women (51). The center of gravity was assumed to be on the line between the joint centers within 40.65 and 41.04% of the thigh and shank lengths for men and within 38.81 and 40.98% for women, respectively (51). The moments were normalized to the BWs of individual subjects. Calculating the absolute moments allowed us to compare the data with those of other studies. The calculations for the mean maximum moments and angles were based on the maximum values for each subject. To allow the calculation of the average repetition within and over all subjects, each individual repetition was sampled into 100 data points. The calculations were performed using MATLAB (version r2010; MathWorks, Natick, MA, USA). Because the moments in the frontal and transverse planes were less reproducible (112), only the sagittal moments are presented. In average over all subjects and load conditions, the maximal knee flexion angle was 85 ± 11° during restricted and 106 ± 10° during unrestricted squats, respectively. The subject who had the smallest knee flexion reached a maximal knee flexion angle of 64°. This occurred during restricted squatting. Therefore, an odd knee flexion angle of 60° in the descent phase because this knee flexion angle was reached by all subjects for all conditions, and the maximum knee flexion angle at the
Figure 1: Experimental setup in the laboratory. A) Motion capture system, B) one force plate for each foot, C) video camera for the sagittal view on the right side (not visible), D) projection of the sagittal view that includes the visible line for the restricted knee motion, and E) line for the restriction (not visible).

deepest position were chosen for the angle and moment comparisons. The measurements were randomly distributed concerning the time of the day and time of year.

**Statistical Analyses**

The influence of the squat technique (restricted vs. unrestricted) and the extra load on the normalized moments of the knee and hip at the maximum angle was analyzed using a multiple repeated-measures analysis of variance. Eight valid executions were averaged for the statistical calculations of each technique. Significance was determined at $p < 0.05$. The partial $\eta^2$ was more than 0.443 for the knee and hip moments and more than 0.667 for the maximal knee flexion angle. The statistical evaluation of the data was
Table 2: The maximal knee (M knee) and hip (M hip) moments and sagittal angles during squatting. *References: Fry et al. (72); Abelbeck (1); Hirata and Duarte (97); Fry et al. (72); Wretenberg et al. (246).

<table>
<thead>
<tr>
<th>Reference</th>
<th>Execution type</th>
<th>M knee (N·m)</th>
<th>M Knee (N·m·kg⁻¹)</th>
<th>M hip (N·m)</th>
<th>M hip (N·m·kg⁻¹)</th>
<th>Knee angle (°)</th>
<th>Hip angle (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fry et al. (14)</td>
<td>Unrestricted</td>
<td>150.1 ± 50.8</td>
<td></td>
<td>28.2 ± 60.0†</td>
<td>66.1 ± 10.0</td>
<td>66.7 ± 6.1</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Restricted</td>
<td>117.3 ± 34.2†</td>
<td></td>
<td>302.7 ± 71.2†</td>
<td>73.4 ± 10.5†</td>
<td>60.6 ± 10.4</td>
<td></td>
</tr>
<tr>
<td>Abelbeck (1)</td>
<td>Foot</td>
<td>~1150</td>
<td></td>
<td>~35</td>
<td></td>
<td>80</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Position 1</td>
<td>~750</td>
<td></td>
<td>~1,100</td>
<td></td>
<td>80</td>
<td></td>
</tr>
<tr>
<td>Hirata and Duarte (19)</td>
<td>T (trespassing)</td>
<td>0.38 ± 0.09§</td>
<td></td>
<td>0.21 ± 0.10§</td>
<td>78 ± 18</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>NT (not trespassing)</td>
<td>0.27 ± 0.03§</td>
<td></td>
<td>0.10 ± 0.05§</td>
<td>92 ± 15</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Wretenberg et al. (40)</td>
<td>Deep squat (powerlifter)</td>
<td>139</td>
<td></td>
<td>324</td>
<td>126 ± 4</td>
<td>146 ± 3</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Parallel squat (powerlifter)</td>
<td>92</td>
<td></td>
<td>309</td>
<td>111 ± 5</td>
<td>132 ± 4</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Deep squat (weightlifter)</td>
<td>191</td>
<td></td>
<td>230</td>
<td>138 ± 3</td>
<td>125 ± 4</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Parallel squat (weightlifter)</td>
<td>131</td>
<td></td>
<td>216</td>
<td>116 ± 5</td>
<td>111 ± 8</td>
<td></td>
</tr>
<tr>
<td>This work</td>
<td>Unrestricted without load</td>
<td>64 ± 17</td>
<td>0.97 ± 0.14</td>
<td></td>
<td></td>
<td>41.4 ± 12.4</td>
<td>0.63 ± 0.14</td>
</tr>
<tr>
<td></td>
<td>Unrestricted + ( \frac{1}{2} ) BW</td>
<td>80 ± 23</td>
<td>1.21 ± 0.20</td>
<td></td>
<td></td>
<td>63 ± 18</td>
<td>0.96 ± 0.20</td>
</tr>
<tr>
<td></td>
<td>Unrestricted + ( \frac{1}{2} ) BW</td>
<td>95 ± 25</td>
<td>1.44 ± 0.20</td>
<td></td>
<td></td>
<td>83 ± 25</td>
<td>1.26 ± 0.29</td>
</tr>
<tr>
<td></td>
<td>Restricted without load</td>
<td>50 ± 15</td>
<td>0.75 ± 0.13</td>
<td></td>
<td></td>
<td>45 ± 13</td>
<td>0.67 ± 0.11</td>
</tr>
<tr>
<td></td>
<td>Restricted + ( \frac{1}{2} ) BW</td>
<td>61 ± 17</td>
<td>0.91 ± 0.16</td>
<td></td>
<td></td>
<td>70 ± 19</td>
<td>1.06 ± 0.17</td>
</tr>
<tr>
<td></td>
<td>Restricted + ( \frac{1}{2} ) BW</td>
<td>71 ± 19</td>
<td>1.07 ± 0.17</td>
<td></td>
<td></td>
<td>95 ± 25</td>
<td>1.44 ± 0.22</td>
</tr>
</tbody>
</table>

*BW = body weight.
†Significantly different from knee moment (p < 0.05).
§Significantly different from unrestricted (p < 0.05).
||Normalized to body mass and body height.
|||Significantly different from other loading conditions at the same execution type.
|Significantly different from restricted.
Figure 2: Average repetitions over all subjects for knee flexion angles (in degrees) vs. knee moments (in Nm kg\(^{-1}\)). Red: 0 BW load, black: 1/4 BW load, blue: 1/2 BW load. The lines represent the restricted squats, and the dots are the unrestricted squats. BW = body weight.
Figure 3: Average repetitions over all subjects for hip flexion angles (in degrees) vs. hip moments (in Nm kg$^{-1}$). Red: 0 BW load, black: 1/2 BW load, blue: 1/2 BW load. The lines represent the restricted squats, and the dots are the unrestricted squats. BW = body weight.
Figure 4: Average repetitions over all subjects for knee flexion angles (in degrees) vs. hip flexion angle (in degrees). Red: 0 BW load, black: 1/4 BW load, blue: 1/2 BW load. The lines represent the restricted squats, and the dots are the unrestricted squats. BW = body weight.

Table 3 The average knee (M knee) and hip (M hip) moments and average hip flexion angles at a knee flexion angle of 60°.

<table>
<thead>
<tr>
<th>Load</th>
<th>Execution type</th>
<th>M knee (N·m·kg⁻¹)</th>
<th>M hip (N·m·kg⁻¹)</th>
<th>Hip angle (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Without load</td>
<td>Unrestricted</td>
<td>0.59 ± 0.10 ‡‡</td>
<td>0.31 ± 0.11 ‡‡</td>
<td>59.7 ± 9.3 ‡</td>
</tr>
<tr>
<td>1/4 BW</td>
<td></td>
<td>0.68 ± 0.15 ‡‡</td>
<td>0.51 ± 0.18 ‡‡</td>
<td>59.9 ± 8.8 ‡</td>
</tr>
<tr>
<td>1/2 BW</td>
<td></td>
<td>0.82 ± 0.16 ‡‡</td>
<td>0.66 ± 0.22 ‡‡</td>
<td>56.7 ± 9.0 ‡</td>
</tr>
<tr>
<td>Without load</td>
<td>Restricted</td>
<td>0.54 ± 0.10 †</td>
<td>0.48 ± 0.13 †</td>
<td>77.4 ± 9.0</td>
</tr>
<tr>
<td>1/4 BW</td>
<td></td>
<td>0.60 ± 0.14 †</td>
<td>0.77 ± 0.16 †</td>
<td>76.2 ± 7.7</td>
</tr>
<tr>
<td>1/2 BW</td>
<td></td>
<td>0.70 ± 0.16 †</td>
<td>1.03 ± 0.17 †</td>
<td>72.9 ± 6.4 †</td>
</tr>
</tbody>
</table>

*BW = body weight.
†Significantly different from the other loads using the same technique.
‡Different from restricted.
performed using IBM SPSS software (version 19; SPSS AG, Zurich, Switzerland).

3.4 Results

In general, additional loads increased the maximum moments in the knees and hips during the restricted and unrestricted squats (Table 2). The maximum knee moments during the restricted squats were 22.4% less than the unrestricted squat moments for 0 BW, 24.6% less for \( \frac{1}{4} \) BW, and 25.4% less for \( \frac{1}{2} \) BW. The maximum hip moments during the restricted squat were 6.9% higher than the unrestricted squat moments for 0 BW, 11.3% higher for \( \frac{1}{4} \) BW, and 14.6% higher for \( \frac{1}{2} \) BW. The maximum moment for the knee was at the lowest point of the squat at the knee flexion angle of approximately 83° for the restricted and approximately 104° for the unrestricted techniques. At the lowest point, the ankle flexion angle is different, 21.3° for the restricted and 31.5° for the unrestricted squat, on average. The moments were different for each load condition (Figure 2). The maximum hip moments for both techniques occurred at hip flexion angles between 95° and 100° (Figure 3). A comparison of the angles in the knee and hip (Figure 4) revealed that the hip flexion angle was larger during the restricted squats. The form of execution had a larger influence on the corresponding hip flexion angle than the extra load. At a 60° knee flexion angle, the moments for the knee were 17% smaller and the moments in the hip were 56% higher for the restricted squats than for the unrestricted squats under a \( \frac{1}{2} \) BW extra load (Table 3). The average ankle flexion angle was different for restricted (12.7°) and for the unrestricted squats (17.9°). The hip flexion angle was approximately 17° larger during the restricted squats. Similar to the deepest position, the moments for the knee and hip at a 60° knee flexion angle were higher with additional loading under both conditions. Furthermore, for these 2 positions, the moment in the knee was higher for the restricted squat and lower
in the hip for unrestricted squats.

The moment in the knee was similar during the descending and ascending squat movements (Figures 2 and 3). The acceleration involved in changing the direction of the movement did not result in a significant increase in the moment at the lowest squat position. The anterior-posterior deviation of the COP during an average repetition with 1/2 BW extra weight was the same for both types of execution (unr: 47 ± 19 mm, r: 50 ± 19 mm). In addition, no difference was observed for the anterior motion of the bar during an average 1/2 BW extra weight repetition (unrestricted squat: 59±24 mm, restricted squat: 50 ± 21 mm).

3.5 Discussion

Considering the complexity of the exercise and the many variables related to performance, understanding squat biomechanics is of great importance both for achieving optimal muscular development and for reducing the prospect of a training-related injury (199). Numerous guidelines based on experience and biomechanical analyses are in use. The main focus has been on the load conditions at the lowest squat position. To our knowledge, this is the first study to compare moments for specific knee flexion angles. On the basis of existing knowledge, we hypothesized that the higher moment in the knee during the squat is derived primarily from the unrestricted knee position. This study demonstrates that the load on the knee at a given knee flexion angle was slightly higher during unrestricted squats, whereas the hip load was clearly lower. The moment in the knee during squatting is mainly given by the angle of the shank (35), the COP, and the direction of the ground reaction force. The increase of the moments in the knee corresponds to the observed larger dorsiflexion of the ankle. The calculated moments for the knee in this work were comparable to those for powerlifters during parallel squats (Table 2) (246). In general, the moments that we observed in the
knee during the restricted and unrestricted squats were lower than those for squats with an extra load equal to 1 BW (72). The greater maximum moment in the knee during unrestricted squats compared with restricted squats occurs at larger knee flexion angles and can be explained by the larger dorsiflexion of the ankle joint that yields larger moment arms for the ground reaction force. The load condition on the knee with an extra load of \( \frac{1}{2} \) BW are comparable to those generated by stair descent (1.3 Nm kg\(^{-1}\) at a knee flexion angle of 60°) (186). Overall, the knee and hip moments were higher for deeper squats. This observation shows the importance of choosing the proper squat depth for each individual. A comparison of the hip moments measured here with those of other studies (Table 2) revealed that the calculated moment in this study was lower than that observed for weightlifters and powerlifters (246) but higher than that observed for 7 weight-trained men (72), who reported a 10-fold increase in the maximum hip moment for restricted squats compared with unrestricted squats. Assuming a similar ground reaction force at the deepest position, this increase of the moment in the hip would increase the external moment arm of the hip by a factor of 10. This large difference in the maximal hip moment between restricted and unrestricted squats was not observed in our study. When focusing on the maximum load and angles during squats, it must be considered that the maximal effort may occur at the deepest point of the squat. However, by comparing the knee and hip moments for certain angles in the knee and hip, the local effects from the form of execution can be studied. The load condition on the knee joint at a given quadriceps length and the corresponding load condition for other parts of the body can thus be determined. In this study, it was observed that the moment in the hip was larger for restricted squats at a knee flexion angle of 60°. Assuming a simple kinematic chain model, the larger observed moments in the hip during restricted squats lead to a higher load on the lower spine. During wide stance back squats, the
lumbar flexion range was 12.6° for women and 25.4° for men (152). The moment arm of the gravitational force of the extra load can be influenced by the position of the barbell on the body. A front squat resulted in lower compression forces and knee extensor moments for the knee (84). However, changing the extra load position will change the relationship of the angles in the knee, hip, and spine. The main load axis of the human spine is along the direction of the spine (158). During squats, this axis receives most of the load in the upright position at the start of the lift. In addition to the knee, the lower back is the region that most commonly experiences problems from squats (162). In a forward dynamics approach, additionally to the axial load, at the lowest point, the moment of the lower spine is affected by the position of the center of mass of the torso and of the bar. A movement resulting in a horizontal shift of the center of mass away from the center of support reduced the stability of the subject. For both types of execution, the anterior displacement of the bar was not different. For the safe execution of a squat, it is important to keep the body’s center of mass and the extra load within the central region of the feet. Regarding this safety precaution, no differences were observed for the restricted and unrestricted execution. The knee and hip flexion angles are crucial for the forward tilt of the spine that is necessary to keep the center of mass within the area of support. Contrary to our hypothesis, the hip flexion angle differs between restricted and unrestricted squats at an angle of 60°, but the maximum hip flexion angle is similar for both techniques. This result could be interpreted to mean that the maximal flexion range of motion of the hip is reached during the unrestricted squat and that it cannot be increased for the restricted squat.
3.6 Practical applications

In this study, the angles and corresponding moments for the knee and hip were determined for restricted and unrestricted squats. As expected, the moment in the knee increases with the angle and load. Although the maximum moment is higher in the knee for the unrestricted squat compared with the restricted squat, the unrestricted squat generated comparable moments at the same knee flexion angles. The maximum hip flexion angle was similar for both forms of execution. The stress on the hip, and likely the lower back, is lower during an unrestricted squat. The squat is primarily a knee and hip extension exercise. The aim is to maximize muscle stress and minimize the load on other parts of the body. Hence, a certain knee moment is necessary to stimulate quadriceps contraction. Restricting the position of the knee necessitates either a shift in the COP toward the heel or a compensatory mechanism in the upper body. A shift in the COP away from the center of the area of support reduces the stance stability, which is especially unfavorable when the subject is lifting heavy weights. Maintaining a stable position at the COP in the supporting area of the feet is crucial for the safety of the athlete. A forward movement of the upper trunk is required to maintain balance and simultaneously align the shins vertically. Maintenance of a normal lordotic torso posture results in significant hip flexion during squats. This movement is contrary to the NSCA guidelines that specify that the torso should be close to vertical during the entire lift. Unrestricted squats are preferable for most athletes because of the higher load for the target muscle with a similar load on the hip and a smaller load of the spine. Great care must be taken with the squat depth, given the increased knee moment and the stress on the hip and lower back at the deepest position.
Acknowledgments

This work was supported by the Eidgenössische Sport Kommission. The technical support from Thomas Ukelo during the measurements is gratefully acknowledged. Alex Stacoff† was an investigator on this project. We miss him.
The aim of this study was to correlate biomechanical parameters during training with the performance of skijumpers during competition. Not only an expected correlation of the vertical take off velocity during training with the personal jump performance was found but also the more the athletes tended toward a valgus knee alignment during the measured movements, the worse their performance. This preliminary results suggest that the strength and conditioning training should also concentrate on improving the knee alignment of the skijumpers.

This paper is in the top 25% of the total research output scored by Altmetric (Wolters Kluwer).

**Contributions:** Within this project, Carole Pauli and Melanie Keller performed their master thesis. SL designed the study, helped with and supervised the data acquisition, evaluation and interpretation. All authors provided text for the final manuscript.
4 Kinematics and kinetics of squats, drop jumps and imitation jumps of ski jumpers

CAROLE A. PAULI1, MELANIE KELLER1,2, FABIAN AMMANN3, KLAUS HÜBNER4, JULIA LINDORFER1, WILLIAM R. TAYLOR1 AND SILVIO LORENZETTI1

1 Institute for Biomechanics, ETH Zurich, Switzerland
2 Department for Sport, Movement and Health, University Basel, Basel, Switzerland
3 Swiss Ski, Haus des Skisportes, Bern, Switzerland
4 Swiss Federal Institute of Sports, Magglingen, Switzerland

Address correspondence to Silvio Lorenzetti, sl@ethz.ch

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4.1 Abstract

Squats, drop jumps, and imitation jumps are commonly used training exercises in ski jumping to enhance maximum force, explosive force, and sport-specific skills. The purpose of this study was to evaluate the kinetics and kinematics of training exercises in ski jumping and to find objective parameters in training exercises that most correlate with the competition performance of ski jumpers. To this end, barbell squats, drop jumps, and imitation jumps were measured in a laboratory environment for 10 elite ski jumpers. Force and motion data were captured, and the influence of maximum vertical force, force difference, vertical take-off velocity, knee moments, knee joint power, and a knee valgus/varus index was evaluated and correlated with their season jump performance. The results indicate that, especially for the imitation jumps, a good correlation exists between the vertical take-off velocity and the personal jump performance on the hill \((R = 0.718)\). Importantly, however, the more the athletes tended toward a valgus knee alignment during the measured movements, the worse their performance \((R = 0.729 \text{ imitation jumps}; R = 0.685 \text{ squats})\). Although an evaluation of the athletes' lower limb alignment during competitive jumping on the hill is still required, these preliminary data suggest that performance training should additionally concentrate on improving knee alignment to increase ski jumping performance.

Key words: Performance, ski jumping, movement analysis

4.2 Introduction

Ski jumping has been an Olympic sport since the winter games of 1924 and it still attracts the attention of spectators and the media alike (164; 200). The introduction of new techniques such as the V-style in 1987 (200), optimized training methods, new materials, and increased training infrastructure over the years have led to large improvements in the performance of athletes, in-
cluding greater jump distances (144). However, although the availability of training facilities has been enhanced, ski jump training on the hill remains extremely time consuming. High-quality performance training in other environments is therefore a key factor for improving competitive performance.

A ski jump is composed of 4 main phases: in-run, take off, flight, and landing (53; 116), although only the first 3 are considered to be essential for jump distance (145). Together with the quality of the skis and their preparation (144), an optimal tucked position during in-run should reduce drag and help increase the jumpers’ take-off velocity (201). After take off, the aerodynamic abilities of the athletes can be decisive for jump distance and therefore the outcome of a competition. However, it is the rapid knee extension during take off, the second phase of the ski jump, that is thought to be the key factor for jump performance in competitive jumping, and one that can also be improved by training strength, timing, coordination, and speed (144).

The take-off velocity on the hill, composed of the in-run velocity and the vertical take-off velocity, reaches its peak approximately 0.3 seconds after the beginning of the take off (117; 163; 191; 231). The explosive strength needed for a large vertical component of the take-off velocity is especially important on smaller hills with jump distances of less than 95 minutes (48), where a high rate of force development increases the take-off velocity (201). The performance training to enhance the take-off typically includes exercises such as squats, drop jumps, and imitation jumps (7; Palazzi and Williams; 236; 240). However, it remains unclear which parameters during these exercises have a primary influence on the performance of the athletes on the hill.

The squat is one of the most popular exercises for performance training to strengthen the muscles of the lower extremities, involving multiple joints and a variety of muscles (1; 71; 136; 142). Although there are no studies to date that have compared the biomechanics of squats with the performance
of ski jumping, it is plausible that training the maximum force in the lower extremities could improve the vertical take-off velocity and with it performance on the hill. Whether it is strength or other factors that are important for jump performance remains to be investigated.

When drop jumping from an elevated platform, immediately followed by a jump over a hurdle, the stretch-shorten cycles of the quadriceps and gluteus maximus are specifically trained. During landing, these muscles are loaded eccentrically, followed by a concentric muscle contraction for the take off (18). Although this exercise is not predominantly sport specific, improvements in jumping skills due to higher muscle force and power output have been shown (18) and therefore the drop jump has been integrated into ski jump training. Bobbert (19) and Walsh et al. (235) came to the conclusion that a sport-specific technique should be chosen for drop jumps in performance training to improve the desired parameters. Importantly, excessive internal or external rotation of the knee resulting from the eccentric-concentric loading cycle can lead to injuries of the passive structures such as the anterior cruciate ligament (18; 95; 96; 173). As it remains unclear which aspects of this training exercise are correlated with the final jumping performance on the hill, it is not yet possible to reduce the highest risk elements without reducing the efficacy of the training.

Although parameters of movement in the sagittal plane have more often been measured during competition than in a laboratory environment (200), 3D parameters of jump take-off kinematics that include e.g., limb alignment during hill jumps are hardly available. Despite differences in comparison with hill jumps, the knee valgus/varus during take off can be evaluated during imitation jumps (200). Starting from a squatting position on the ground, the athletes simulate the take-off jumping action and are held in the air by the trainer to mimic the actual take-off motion on the hill as closely as possible (229). To better understand the process, Virmavirta and
Komi (232) investigated the kinetics of imitation jumps in a laboratory environment. Although the largest forces were indeed observed in the vertical direction, as would be expected for a large take-off velocity, forces in antero-posterior direction were also observed for all athletes, which is not possible on the hill due to the low friction between the skis and in-run track. In addition, Müller (164) suggested a sufficiently high vertical take-off velocity of at least 2.5 m/s, but further increases in the take-off velocity, which can only be achieved by extreme effort, might be less important than optimized take-off movements (164). The force during take off should be applied vertically (116; 229) and symmetrically for an effective take off. The knee is seen as the joint with the highest power production during jumps (192). However, which kinetic and kinematic parameters during training exercises correlate with performance on the hill remain unknown. An evaluation of those parameters in the training of elite ski jumpers and an associated correlation with the performance is required for a reduction of injury risk and an optimal focus on the most decisive parameters for ski jumping performance in strength and jumping exercises.

To this end, the objective of this study was to determine kinetic and kinematic parameters of ski jumpers during performance training and to correlate these biomechanical parameters with their jump performance during competitions in the summer season 2012. Especially the maximum vertical force, force difference between the legs, knee valgus/varus index, and joint moments in the knee of ski jumpers were analyzed during squat. In addition to these parameters, the vertical take-off velocity and joint power in the knee were determined during drop jumps and imitation jumps. With regard to earlier studies of biomechanical parameters in this field and expertise in ski jumping, it is hypothesized that normalized kinetic parameters maximum force, knee moments and power, as well as the vertical take-off velocity and the knee valgus/varus index, correlate positively with the ath-
letes’ performance, whereas lower force differences between the legs result in better performance and therefore show a negative correlation with jumping performance.

4.3 Methods

Experimental Approach to the Problem

Each athlete performed 2 sets of squats and 1 set of drop jumps and imitation jumps. The extra load on the barbell during squats corresponded to the athletes’ actual training weight and 70% of their 1 repetition maximum (1RM). Ground reaction forces and motion data were used to analyze the maximum vertical force, force differences between the legs with respect to the maximum vertical force, the maximum knee joint moments, and the knee valgus/varus index during all 3 exercises, and the maximum vertical take-off velocity and knee joint power during drop jumps and imitation jumps. The knee valgus/varus index was analyzed in terms of minimum values and their values at the maximum knee flexion angle. The evaluated biomechanical parameters were determined at the beginning of the winter season and correlated with the ski jumping performance in competitions during the previous summer season 2012. All measurements were completed during a single visit at the movement analysis laboratory of the Institute for Biomechanics (IIB) at the ETH Zurich.

Subjects

The subjects in this study represented the top end of ski jumpers in Switzerland. Here, 1 female and 9 male subjects with a mean age of 23 ± 4 years (range, 19–31), an average height of 179 ± 5 cm, and an average weight of 64.6 ± 4.8 kg participated in this study. The 7 elite ski jumpers and 3 elite Nordic combined (ski jumping and cross-country skiing) athletes were all members of the national performance center of the Swiss Ski Federation in
Einsiedeln (Switzerland), and were all experienced in strength training. All athletes were free of injuries and health problems at the time of the study. The study was approved by the Ethics Committee of the ETH Zurich, and written informed consent to participate in the study was obtained from all subjects after receiving detailed information about the measurement procedures.

Procedures

Kinetic data was measured using 2 Kistler force plates (Type 9286AA; Kistler Instrumente AG, Winterthur, Switzerland) with a sampling frequency of 2000 Hz (7). An optoelectronic measurement system (Vicon V612; Oxford metrics, Oxford, United Kingdom) with 12 cameras (MX40; 8 fixed, 4 mobile; resolution 2353 × 1728 pixels) (2) and a sampling frequency of 100 Hz was used to capture the motion during the exercises. Seventy-seven skin markers based on the IfB marker set of List et al. (136), with 6 additional markers on the arms were then fixed to the subjects by the same examiner.

After an individual warm-up and the equipping with the skin markers, the measurements including squats, drop jumps, and imitation jumps were conducted. The first set of squats composed of 5 repetitions and an extra load corresponding to the subjects’ actual training weight was followed by a set of 5 repetitions with an extra load of 70% of the estimated 1RM of each athlete (Table 1). The 1RM was estimated as follows: First, an isometric maximum force with maximum voluntary contraction (MVC) test for the squat position was performed at the Swiss Federal Institute of Sports Magglingen, Switzerland. This test is part of the typical performance diagnostics for ski jumpers and is conducted on a regular basis during their noncompetitive phase. Compared with 1RM testing, this approach ensures a higher safety standard and lower risk of injury. Subjects pushed maximally against a bar, fixated at a 70° ski jump–specific knee angle position,
which was controlled by a goniometer. The subjects’ feet were placed on a force plate (MLD Test Evo 2; SPSport, Innsbruck, Austria). Total ground reaction force (sum of both legs) and knee angle data were collected and saved in a database. The conversion factors for MVC to 1RM of 71.3% for male and 67.1% for female subjects were based on a study by Duss and Hobi (7), who investigated the correlation between MVC and 1RM in different knee angles for 12 male and 7 female highly-trained ski alpine athletes. Instructions given for the squats were similar to a previous study conducted at the Institute (19) (Table 2). As a measure of reproducibility, the typical coefficient of multiple correlation values for the lower extremity motion during squatting were about 0.97 (sagittal plane) and 0.8–0.85 (frontal-/transverse plane) (18).

The subjects performed the drop jumps starting from an upright position on a platform with a height of 74 cm, with the tips of their shoes flush with the platform edge. They were instructed to drop from the platform and

<table>
<thead>
<tr>
<th>Subject No.</th>
<th>First set training weight</th>
<th>Second set 70% 1RM</th>
</tr>
</thead>
<tbody>
<tr>
<td>S01</td>
<td>90</td>
<td>80</td>
</tr>
<tr>
<td>S02</td>
<td>80</td>
<td>80</td>
</tr>
<tr>
<td>S03</td>
<td>90</td>
<td>75</td>
</tr>
<tr>
<td>S04</td>
<td>93</td>
<td>93</td>
</tr>
<tr>
<td>S05</td>
<td>90</td>
<td>93</td>
</tr>
<tr>
<td>S06</td>
<td>85</td>
<td>85</td>
</tr>
<tr>
<td>S07</td>
<td>85</td>
<td>92.5</td>
</tr>
<tr>
<td>S08</td>
<td>90</td>
<td>80</td>
</tr>
<tr>
<td>S09</td>
<td>95</td>
<td>87.5</td>
</tr>
<tr>
<td>S10</td>
<td>70</td>
<td>72.5</td>
</tr>
</tbody>
</table>
Table 2: Instructions for execution of squats Lorenzetti et al. (140).

<table>
<thead>
<tr>
<th></th>
<th>Instructions</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Stand upright with your feet approximately shoulder width apart</td>
</tr>
<tr>
<td>2</td>
<td>Point the feet slightly outward, following the natural divergence of the feet</td>
</tr>
<tr>
<td>3</td>
<td>Put the barbell on the trapezius muscle and hold it with a comfortable hand position</td>
</tr>
<tr>
<td>4</td>
<td>Lift the thorax to a natural spinal position</td>
</tr>
<tr>
<td>5</td>
<td>Hold tension in the core muscles during execution of the squat</td>
</tr>
<tr>
<td>6</td>
<td>Breathe out during the ascent</td>
</tr>
<tr>
<td>7</td>
<td>Perform the squat explosively</td>
</tr>
</tbody>
</table>

immediately rebound over a hurdle (whose height and distance they were free to choose) while keeping the ground contact time as short as possible. For all subjects, the hurdle was higher than the dropping height. Six valid trials were required from each athlete, which involved both feet landing completely on the force plates and no loss of markers before clearance of the hurdle.

Finally, 10 imitation jumps without additional weight were conducted with the help of a trainer, in the same manner as they are executed during training. The subjects were allowed sufficient, individually chosen rest periods between trials.

**Cycle Definition**

Start and end time points of the movements were defined for all 3 exercises as follows. A squat repetition was defined as starting in an upright position, moving down to the lowest position achieved during the squat and returning upward to the original posture. The start and end points of the cycle were defined by the vertical velocity of the barbell ($v_{barb} < 0.04 \text{ m/s}$) tracked by
2 markers attached to the ends of the barbell (136).

The evaluation of the knee valgus/varus index at the position of the maximum knee angle of the drop jumps was determined during the movement starting when the data from at least 1 force plate had exceeded 2% of the subject’s body weight (BW). Accordingly, the end point occurred when the force on both plates was again lower than 2% BW. The remaining variables were determined from the starting point of the drop jump defined as the lowest crouched position of the athletes, derived from the average of the markers fixed to the acromia, to the end point derived from the data of the force plates as described above.

Finally, the imitation jumps started from where the take-off velocity, calculated from force data, became and remained > 0 until the point when the maximum velocity was reached.

**Kinetics**

The maximum vertical force for the stance phase of each repetition of the different exercises was determined as the sum of forces for both legs, normalized to BW. As a measure of asymmetry, the force production from both legs was evaluated using the absolute difference in maximum vertical force for each foot as a percentage of the total maximum vertical ground reaction force (equation 1).

$$\Delta F_{\text{max}} = \frac{|F_{\text{max,left}} - F_{\text{max,right}}|}{F_{\text{max}}} \cdot 100$$  \hspace{1cm} (1)

where $\Delta F_{\text{max}}$ [%$F_{\text{max}}$] is the difference between maximum vertical force of the left and right foot; $F_{\text{max,left}}$ [N] is the maximum vertical force under the left foot; and $F_{\text{max,right}}$ [N] is the maximum vertical force under the right foot. The maximum of the vertical component of the take-off velocity for drop jumps and imitation jumps was determined from the vertical force data (equation 2):
\[
v(t) = \int_{t_{\text{start}}}^{t_{\text{end}}} \frac{F_{\text{GRF}} - F_g}{m} dt
\]  

(2)

where \( v \) [m/s] is the velocity of the subject’s center of mass; \( F_{\text{GRF}} \) [N] is the vertical ground reaction force; \( F_g \) [N] is the bodyweight; and \( m \) [kg] the subject mass. Knee joint moments (normalized to BW) were calculated using functionally determined joint centers from basic motion tasks (136). The maximum values are derived from the average of both legs. The normalized joint moments \( M \) [Nm/kg] and the joint angular velocity \( \omega \) [1/s] in the knee, also reported as the average of both legs, were combined to calculate the normalized maximum of the joint power, \( P \) [W/kg] in the knee (equation 3).

\[
P = M \cdot \omega
\]  

(3)

**Kinetics**

The index for knee valgus/varus, \( \Delta d^* \), was calculated using equation 4:

\[
\Delta d^* = \frac{k - a}{a}
\]  

(4)

where \( k \) is the distance between knee joint centers; and \( a \) is the distance between ankle joint centers. \( \Delta d^* = 0 \) indicates straight leg alignment, whereas \( \Delta d^* < 0 \) indicates a knee valgus and \( \Delta d^* > 0 \), a knee varus. The distances \( k \) and \( a \) were assessed at the lowest body position (i.e., largest knee flexion angle) during the squat and drop jump exercises (\( \Delta d_{\text{knee}}^* \)), similar to the study of Herrington and Munro (95). Starting from a crouched position at the maximum knee flexion angle during the imitation jumps, was calculated as the average of \( \Delta d^* \) during the first 10% of the exercise. Differing from (95), the distances were derived from 3D motion data and were therefore an extension of a planar analysis. Additionally, the lowest value of \( \Delta d^* \)
during the execution of the 3 exercises was evaluated ($\Delta d_{\text{min}}^*$). The joint centers were calculated from the marker data, where normal standing was considered as the reference for a neutral posture, defining the knee angle of $0^\circ$. The angle was then calculated as the relative motion of the lower limb relative to the upper limb.

**Performance**

The evaluation of ski jumping performance was based on the points achieved during a competition season. To directly compare different competitions under different environmental conditions, expertise and weighting was required, as shown in major sport rating systems (208). Similar to the calculation of alpine FIS points (FIS), Swiss Ski (214) uses a scoring table (Table 3) that has been adapted thoroughly by ski jumping trainers over the years. The ranking is based on the average of the best 6 Swiss Ski points during a competition period, where the points are calculated as follows: The points achieved during a competition, resulting from jump length and the judges’ notes, are divided by 10 and weighted for different competition types using specific extras and deductions (Table 3). This scoring method was used in the present study for evaluating the jumping performance of each athlete on the hill in the summer season of 2012, which was shortly before the investigation was conducted.

**Statistical Analyses**

The mean values and SDs over the valid trials were determined for each parameter and each athlete. The Shapiro-Wilk method was used to test the trials of each athlete and the parameters as the averaged values of the athletes for normal distribution.

As this could be shown for 90% of the tests, normal distribution was adopted for all parameters. A correlation analysis was conducted for the
Table 3: Calculation of points for international competitions SwissSki (213).

<table>
<thead>
<tr>
<th>International competitions</th>
<th>Points</th>
</tr>
</thead>
<tbody>
<tr>
<td>World Cup (Men), World Championships, Olympic games</td>
<td>+6</td>
</tr>
<tr>
<td>Junior World Championships, Continental Cup (Men)</td>
<td>+4</td>
</tr>
<tr>
<td>FIS-Cup/Alpen-Cup</td>
<td>+2</td>
</tr>
<tr>
<td>World Cup (Women)</td>
<td>0</td>
</tr>
<tr>
<td>Continental Cup (Women)</td>
<td>−1</td>
</tr>
<tr>
<td>FIS-Cup/Alpen-Cup (Women)</td>
<td>−3</td>
</tr>
</tbody>
</table>

Example: Continental Cup
Achieved points in competition (2 jumps) 247.5 Points
Division by 10 24.75 Points
Points for competition (+4) 28.75 Points
parameters $F_{\text{max}}$, $\Delta F_{\text{max}}$, $v$, $\Delta d^*$, $M_{\text{max}}$, $P_{\text{max}}$ with the jumping performance. The IBM software package SPSS Statistics version 22 (IBM Corp., Armonk, NY, USA) was used for all analyses with an alpha level of 5% ($p \leq 0.05$).

4.4 Results

Kinetics

Within 1 set, most of the athletes conducted the squats regularly with a relative SD of less than 5%, whereas some varied up to 18% with maximum forces of $38.7 \pm 7.7$ N/kg (first set) and $39 \pm 3.8$ N/kg (second set) (Figure 1). While performing drop jumps, maximum forces of $49 \pm 6.3$ N/kg were shown with a relative SD below 7% for 90% of the athletes. The imitation jumps, as a sport-specific exercise, were conducted very regularly, resulting in low relative SDs for the individual subjects (¡2% for 80% of the athletes) and an average maximum force of $24.6 \pm 2.5$ N/kg.

The interlimb force variability, $\Delta F_{\text{max}}$ for the imitation jumps was below 1% for some of the subjects, whereas others exhibited interlimb differences of up to 3% (Table 4). Similar values were found for the drop jumps, but mean force differences between the legs of up to 11% were shown for the squats.

The highest knee joint moments were achieved during the drop jumps. Relative SDs of up to 16% for squats and drop jumps are in contrast to values of less than 5% for all athletes during the imitation jumps (Table 5). Similar to the maximum moments, the maximum knee joint power was higher during drop jumps than during imitation jumps.

Taking advantage of the stretch-shortening cycle during drop jumps, average vertical velocities of $3.35 \pm 0.30$ m/s were achieved compared with $2.95 \pm 0.23$ m/s during imitation jumps.
Figure 1: Maximum vertical forces $F_{max}$ (N/kg)—normalized mean and SD for all subjects (squats—top, drop jumps—middle, imitation jumps—bottom).
Table 4: $\Delta F_{\text{max}} [\% F_{\text{max}}]$—mean and SD for all subjects.

<table>
<thead>
<tr>
<th>Subject No.</th>
<th>Squats</th>
<th>Drop jumps</th>
<th>Imitation jumps</th>
</tr>
</thead>
<tbody>
<tr>
<td>S01</td>
<td>7.9 ± 8.1</td>
<td>2.8 ± 1.9</td>
<td>3.1 ± 0.6</td>
</tr>
<tr>
<td>S02</td>
<td>9.5 ± 6.1</td>
<td>1.4 ± 0.9</td>
<td>1.2 ± 0.5</td>
</tr>
<tr>
<td>S03</td>
<td>10.9 ± 6.5</td>
<td>2.2 ± 1.2</td>
<td>2.0 ± 0.6</td>
</tr>
<tr>
<td>S04</td>
<td>8.0 ± 5.8</td>
<td>3.2 ± 1.6</td>
<td>1.0 ± 0.7</td>
</tr>
<tr>
<td>S05</td>
<td>8.3 ± 5.3</td>
<td>1.0 ± 0.6</td>
<td>1.2 ± 0.8</td>
</tr>
<tr>
<td>S06</td>
<td>9.4 ± 6.3</td>
<td>1.4 ± 1.2</td>
<td>1.3 ± 0.5</td>
</tr>
<tr>
<td>S07</td>
<td>8.0 ± 4.0</td>
<td>2.5 ± 1.5</td>
<td>0.9 ± 0.4</td>
</tr>
<tr>
<td>S08</td>
<td>9.5 ± 7.6</td>
<td>1.7 ± 1.1</td>
<td>1.0 ± 0.5</td>
</tr>
<tr>
<td>S09</td>
<td>7.4 ± 6.6</td>
<td>2.1 ± 1.3</td>
<td>3.0 ± 1.3</td>
</tr>
<tr>
<td>S10</td>
<td>9.1 ± 12.0</td>
<td>0.7 ± 0.6</td>
<td>2.6 ± 1.0</td>
</tr>
<tr>
<td>Mean ± SD</td>
<td>8.8 ± 1.1</td>
<td>1.9 ± 0.8</td>
<td>1.7 ± 0.9</td>
</tr>
</tbody>
</table>

Table 5: Maximum moments $M$ and power $P$ in the knee for drop jumps and imitation jumps.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Squats</th>
<th>Drop jumps</th>
<th>Imitation jumps</th>
</tr>
</thead>
<tbody>
<tr>
<td>$F_{\text{max}}$ (N·m·kg$^{-1}$)</td>
<td>2.61 ± 0.38</td>
<td>6.18 ± 0.58</td>
<td>3.21 ± 0.49</td>
</tr>
<tr>
<td>$P_{\text{max}}$ (W·kg$^{-1}$)</td>
<td>=</td>
<td>52.35 ± 10.36</td>
<td>39.09 ± 7.77</td>
</tr>
</tbody>
</table>

**Kinematic**

At the lowest point of the squats, 90% of the athletes showed a tendency toward knee varus (Table 6), which was in accordance with the anatomical conditions (58). Although the mean value during imitation jumps indicated a knee varus alignment at maximum knee angles, a tendency toward knee valgus was found for 60% of the athletes. However, the average minimum $\Delta d^3$ was negative for all 3 exercises. Within the imitation jumps, this shows a knee valgus alignment during take off. Similarly, this was also the case for the drop jumps, for which the average values at the lowest position in the exercise exhibited a knee valgus alignment.

**Performance**

Each athlete’s jumping performance during the summer season of 2012 was available as a basis for the statistical evaluation (Figure 2).
Table 6: Standardized instructions for squat performance.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Squats</th>
<th>Drop jumps</th>
<th>Imitation jumps</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\Delta \sigma^*_\text{knee}$</td>
<td>$0.14 \pm 0.09$</td>
<td>$-0.12 \pm 0.20$</td>
<td>$0.02 \pm 0.11$</td>
</tr>
<tr>
<td>$\Delta \sigma^*_\text{min}$</td>
<td>$-0.12 \pm 0.08$</td>
<td>$-0.21 \pm 0.15$</td>
<td>$-0.22 \pm 0.11$</td>
</tr>
</tbody>
</table>

Figure 2: Ski jumping performance during the summer season of 2012 for each subject SwissSki (214).
Table 7: Standardized instructions for squat performance.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Squats</th>
<th>Drop jumps</th>
<th>Imitation jumps</th>
<th>Drop jumps</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$r$</td>
<td>$p$</td>
<td>$r$</td>
<td>$p$</td>
</tr>
<tr>
<td>$F_{max}$ (N·kg$^{-1}$)</td>
<td>0.592</td>
<td>0.072</td>
<td>-0.179</td>
<td>0.620</td>
</tr>
<tr>
<td>$\Delta F_{max}$ (N·m)</td>
<td>0.125</td>
<td>0.730</td>
<td>0.110</td>
<td>0.763</td>
</tr>
<tr>
<td>$v_{max}$ (m·s$^{-1}$)</td>
<td>-</td>
<td>-</td>
<td>0.647*</td>
<td>0.043</td>
</tr>
<tr>
<td>$\Delta F$ (knee)</td>
<td>0.502</td>
<td>0.139</td>
<td>0.570</td>
<td>0.085</td>
</tr>
<tr>
<td>$\Delta F$ (min)</td>
<td>0.685*</td>
<td>0.029</td>
<td>0.555</td>
<td>0.121</td>
</tr>
<tr>
<td>$M_{max}$ (knee) (N·m·kg$^{-1}$)</td>
<td>0.639*</td>
<td>0.050</td>
<td>0.318</td>
<td>0.370</td>
</tr>
<tr>
<td>$P_{max}$ (knee) (W·kg$^{-1}$)</td>
<td>-</td>
<td>-</td>
<td>0.607</td>
<td>0.063</td>
</tr>
</tbody>
</table>

*Significant correlation for squats, drop jumps, and imitation jumps.

Correlations

Although the vertical take-off velocities were indeed correlated with the athletes' jumping performance on the hill ($r = 0.647$ for the drop jumps and $r = 0.718$ for the imitation jumps), no significant correlation could be found for the maximum vertical forces within the exercises (Table 7). The highest correlation with the jumping performance was shown in the minimum $\Delta d^*$ during the imitation jumps ($r = 0.729$). Similarly, jumping performance was correlated with the performance during squats ($r = 0.685$), whereas no significant correlation could be shown for the drop jumps in minimum $\Delta d^*$.

Although the maximum knee moments during squats are slightly correlated with ski jumping performance on the hill, no significant correlation could be found for the remaining parameters within the exercises.

4.5 Discussion

Coaches often rely on experience when providing feedback to ski jump athletes during performance training, where take-off force and take-off velocity are commonly measured during exercises to provide a key training goal. For the first time, this study reveals a connection between the take-off velocity and the athletic performance, but at the same time does not find any link with take-off force. However, the investigation further found that the vertical take-off velocity was not the only important parameter: knee val-
gus during squats and imitation jumps seem to be highly correlated with performance on the hill. This effect is likely to be related to the efficiency with which power can be transferred at take off, suggesting that take-off technique can play a more important role for performance than raw power alone. This information can help coaches to know where to focus during training sessions to maximize the benefits of an athlete’s jump training.

Separating the push-off propulsion phase from the landing during drop jumps allows the comparison of biomechanical parameters during take off with other sport-specific exercises. Interestingly, using the average weight of Viitasalo’s et al. (227) triple jumpers to calculate the normalized vertical forces in the propulsion phase of a drop jump from a box 6 cm higher than the one used in the present study suggests that the maximum values are lower than those achieved by the ski jumpers. In contrast, Virmavirta and Komi (229) measured imitation jumps in the laboratory of 10 Finnish elite jumpers. The relative SDs of their maximum forces were clearly higher than those observed in our study. Their measurements on the hill (230) yielded forces that were slightly below those measured in our investigation, but this could be due to the technical set up, including the lack of slope in the laboratory.

As a reflection of performance-determining parameters during take off, the early flight phase is seen as a crucial phase in ski jumping (228). To avoid a leaning position during the early flight phase, it is known that athletes should push with approximately the same force for both legs during take off. Because the subjects in this study represent the top end of ski jumpers, only small differences in the maximum force between the left and right legs were expected, especially for the imitation jumps. The results of this study indicate that if additional load acts on the athlete during the squatting or drop jump training, either from the weight of the barbell or because of the drop from the box, subjects have difficulties balancing the
force on both legs. In contrast, the imitation jumps can be well prepared for, and the athlete can choose the moment of take off and there is no additional load. However, even under these conditions, some athletes had problems to distribute the force equally. It seems that despite requiring a bilateral movement in their sport, ski jumpers still develop a dominant leg under some conditions. This supports the findings in (165), where significant strength imbalances between the dominant and nondominant leg could be shown, even for bilateral movements. As a remaining question in Newton et al. (165), whether imbalances result from sport-specific training or other parameters, the present study showed that the interlimb force variability as a similar parameter to the calculated imbalance in Newton et al. (165) was lower for sport-specific exercises than for common training exercises.

Jumpers in this study achieved average vertical take-off velocities during drop jumps that were higher than those presented by Walsh et al. (235). In agreement with Schwameder et al. (202), the imitation jumps resulted in an average vertical take-off velocity of 2.95 ± 0.23 m/s. According to Virmavirta et al. (229), only approximately 72–85% of this velocity can be applied during jumps on the hill. The resulting velocity of 2.12–2.51 m/s on the hill is close to the 2.5 m/s, which according to Müller (164), is necessary for a good jump distance and is a sign for good explosive force. The only other study that measured imitation jumps showed lower take-off velocities (229). Especially when wearing jumping boots during imitation jumps, only 91.6% of the take-off velocity could be achieved compared with imitation jumps when wearing training shoes (229). Furthermore, these authors summarized that the take-off velocity of jumps on the hill ranged from 2.33 ± 0.10 m/s (233) up to 2.85 m/s (Jost et al.), depending on the size of the hill and the ability level of the ski jumpers (109). Because of the fact that the take-off velocities for hill jumps are similar to imitation jumps, even with different shoes, it seems that the take-off situation during
training is indeed similar to the real jumps, although the slope and the lift are generally missing in laboratory environments.

As a result of the additional loads used in this study, which exceeded the BW of the athletes, the knee moments during squatting were higher than the moments reported in (140) for restricted and unrestricted squats. Slightly lower average values were found for the push-off phase during drop jumps in (20), providing normalized values for the maximum moments in the knee. However, it must be noted that the platform from which the drop jumps were performed was 10 cm lower than in our study. Although kinetic parameters seem to be essential in ski jumping performance, joint moments during hill jumps were evaluated based on motion data (192) and show differing results to the present study. Similarly, the knee joint power during imitation jumps in our study differs from the findings of Sasaki et al. (192), as their values, derived from hill jumps, are markedly higher. The calculation of the kinetic parameters from video data in Sasaki et al. (192) might explain these differences and can therefore not be seen as an appropriate comparison with analyses of imitation jumps in a laboratory environment, for which kinetic data are measured accurately using force plates.

Performing exercises correctly is a key factor for safety in strength training (250). In fitness centers in Switzerland, 21.1% of injuries are due to incorrect execution of the exercise and 45.6% to overloading (162). The alignment of the lower limbs can be seen as one factor critical for correct execution. A straight leg axis or a tendency to knee varus for the lowest position during squats and the starting point during imitation jumps (Table 5) seems plausible. When flexing the knee, an internal rotation of the hip joint leads to a valgus alignment (96). If the foot is not fixed, it is normal that a tibial internal rotation occurs during knee flexion, whereas an external rotation occurs during knee extension (61). During the execution of squats,
however, since the foot is fixed, there is an external rotation of the femur when flexing the knee, which also causes the tibia to rotate externally (61). Despite these anatomical relationships, this study could show that during squats as well as drop jumps and imitation jumps, the knees tend toward a valgus position within the exercises. The trend for a knee valgus during the drop jumps might be explained by the short time required for landing and thus insufficient time for the musculature to react appropriately (96). The leg axis during imitation jumps that shows knee valgus alignment during take off, seems to result from a lack of focus or an inability to control the limb axis, which would support the findings of Wallace et al. (234), who suggested that a lack of neuromuscular control could be the reason for knee valgus position.

List et al. (136) and Lorenzetti et al. (140) looked at the impact of anteroposterior knee motion during unrestricted and restricted executions of squats. If the knee motion is restricted, there is a reduced load on the lower extremities but more load on the lower back, than for unrestricted performance. Therefore, it makes sense that ski jumpers do not restrict their knee movement during take off as the effect on the muscles of the lower extremities will be greater than with restricted squats and the load on the lower back will be lower.

The scale used to rate the ski jump performance during the best 6 competitions of the summer season (214) not only includes the personal performance but also a correction factor, which is based on the type of competition. This factor allowed the comparison of performance throughout an extended period and effectively reduced the influence of single exploits. As a required comparison of different competition types, this method was seen as appropriate for the analysis of ski jumping performance, as it includes, among others, ranks of jump length that might be markedly different between the compared competitions.
The results of this study indicate that the maintenance of limb alignment during take-off during the squat and imitation jump training exercises is correlated with ski jumping performance and should therefore be a dominant parameter for the efficacy of the take-off training in ski jumpers (Table 7). Thus, it seems plausible that knee valgus should be avoided not only at the lowest point of, but throughout the whole training exercises. The magnitude of the knee valgus/varus index for the minimum values indicates that a knee valgus alignment should be noticeable by eye during the training exercises.

Our data support the findings of Müller (164), who suggested a non-linear behavior of the force-velocity relationship, including little improvement in the velocity for further increases in maximum force, and that only a certain amount of maximum force is needed. A slight correlation between the maximum vertical forces and the corresponding jump lengths on the hill was shown in Virmavirta and Komi (230), whereas we did not observe any significant correlation for a score of the ski jumping performance and therefore this hypothesis has been rejected. However, considering take-off velocity and its effect on jump length, we were able to support the findings of Müller (164) by showing a significant correlation between vertical take-off velocity and the ski jumping performance on the hill. Although the effect of imbalances and the hypothesized relation to ski jumping performance could not be shown in the present study, it seems important for the imitation jumps to exert an equal force with both legs during the take-off. Deviations may not have a direct effect in the laboratory or during training, but if the athletes are not balanced during take-off, they may have to correct their position during the early flight phase and therefore not reach stable flight as quickly, or even be forced into a less stable or less aerodynamic position during the early phases of flight until correction can occur. Besides the significant correlation of the normalized moments in the knee during squats, no further relation between the power and moments in the knee with the performance
could be shown. As these parameters are assumed to be crucial in ski jumping, the magnitude of the differences between elite athletes might be too small for significant correlations. Furthermore, a multivariate interaction of all decisive parameters should be considered.

The parameters that were not significantly correlated with performance in this study, but have been seen as important factors in the literature, should be considered during training exercises as they are a basis for the take off, which further influences the phases of flight during ski jumping. The hypothesis was rejected for the evaluated parameters except the vertical take-off velocity, minimum knee valgus/varus index and the normalized moments in the knee for the mentioned exercises. However, it is assumed that the parameters are related to performance if lower level athletes would be considered. In addition to the underlying generic ability of athletes to reach elite level, it is therefore reasonable that subject-specific characteristics and training focus on e.g., limb alignment are required for practicing and perfecting jumping performance.

4.6 Practical Application

Squats, drop jumps, and imitation jumps conducted by ski jumpers were biomechanically analyzed in this study. The maximum vertical forces, force differences, vertical take-off velocities, as well as the knee valgus/varus index, the maximum knee moments and joint power and their correlation with the performance on the hill were calculated. The results not only indicate that it is essential to have a good force basis, but also that a high vertical take-off velocity seems to be much more important than the maximum vertical force. If the athlete shows a knee valgus during the take off, the force can probably not be converted optimally into a high take-off velocity. This would explain why the knee position during take off in the imitation jumps has the highest correlation with the performance on the hill. One
reason for a knee valgus can be instability in the knee joint. For trainers and athletes, this means that proper knee alignment is important during performance training and the magnitude of the values indicates that valgus alignment can be monitored during training without the requirement of motion capture technology. Although no significant correlation of the interlimb force variability for the sport-specific exercise imitation jumps with the performance could be shown, this parameter seems to be crucial for an optimal early flight phase and should therefore not be neglected during performance training. As the top end of ski jumpers was measured and no significant correlation with the most parameters could be found, this indicates that ski jumping performance cannot be estimated by performance in training exercises at the elite level. Even though those parameters are the basis of the take off and provide reference values for coaches, measurements during the time-consuming training on the hill and the athletes’ abilities in the flight phases seem to be required for the evaluation of ski jumping performance. Another parameter that should not be forgotten is the force difference between the left and right leg. This may not have a direct impact on the performance for the squats and drop jumps but more on imitation or hill jumps as sport-specific exercises. To enhance performance during competition (and to reduce the injury risk during training), trainers should ensure the correct execution of all exercises during performance training.
Injuries during weight lifting most commonly occur to the spine. In this paper, the kinematics of the spine during two different type of squats were analyzed to improve guidelines for exercise execution for therapists and coaches. A newly developed marker set, validated using open MRI, was adopted to capture the movement of the spine during squatting. Unrestricted squatting resulted in a larger movement of the upper back.

This paper is in the top 5% of all research output scored by Altmetric (Wolters Kluwer).

The findings of the present work related to squatting were published in a popular non-scientific journal to promote the results amongst strength training practitioners (Lorenzetti, S. 2013, Bewegung und Belastung bei der Kniebeuge, Fitness Tribune, 141, 54-55.

Contributions: Within this project, Turgut Gülay and Mirjam Stoop performed their master thesis. SL designed the study, supported and supervised data acquisition, evaluation and interpretation. All authors provided text for the final manuscript. Alex Stacoff was in an investigator in an early stage of this project.
5 Kinematics of the trunk and the lower extremities during restricted & unrestricted squats

RENATE LIST, TURGUT GÜLAY, MIRJAM STOOP AND SILVIO LORENZETTI

Institute for Biomechanics, ETH Zurich, Switzerland

Address correspondence to Silvio Lorenzetti, sl@ethz.ch

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5.1 Abstract

Squatting is a common strength-training exercise used for rehabilitation, fitness training and in preparation for competition. Knowledge about the loading and the motion of the back during the squat exercise is crucial to avoid overuse or injury. The aim of this study was the measurement and comparison of the kinematics of the lower leg, trunk and spine during unrestricted and restricted (knees are not allowed beyond toes) squats.

A total of 30 subjects performed unrestricted and restricted barbell squats with an extra load of 0%, 25% and 50% bodyweight. Motion was tracked using a 12-camera Vicon system. A newly developed marker set with 24 trunk and 7 pelvic markers allowed us to measure 3D segmental kinematics between the pelvic and the lumbar regions, between the lumbar and the thoracic segments and between the sagittal curvatures of the lumbar and the thoracic spine.

In an unrestricted squat, the angle of the knee is larger and the range of motion (ROM) between the lumbar and the thoracic segments is significantly smaller compared with a restricted squat \( (p < 0.05) \). The studied subjects showed significantly increased ROM for thoracic curvature during restricted squats.

The unrestricted execution of a squat leads to a larger ROM in the knee and smaller changes in the curvature of the thoracic spine and the range of smaller segmental motions within the trunk. This execution in turn leads to lower stresses in the back. To strengthen the muscles of the leg, the unrestricted squat may be the best option for most people. Thus, practitioners should not be overly strict in coaching against anterior knee displacement during performance of the squat.

Keywords: movement of the spine, marker set, joint angles
5.2 Introduction

The squat exercise is a basic movement used in fitness training, strength training and rehabilitation. In the bfu report 39 (162), the squat exercise was found to be one of the most predominant exercises in terms of risk of injury and risk of complaint. The report, assessing the injuries and discomforts that arose during a period of 12 months within a group of 6680 gym members, found that leg presses and free-weight squat exercises lead to back injuries and back discomfort (162). The proper form of execution of the squat exercise is essential to minimize its injury risk (32).

The squat can be performed in a restricted (r) or an unrestricted (unr) manner. In the r type squat, the shanks are only allowed to move until the most anterior part of the knees reach a vertical line extending upward from the toes. The r type squat is frequently used in fitness centers (32). A 1970s study (6) stated that greater shearing forces impinge upon the knee in unr type squats and as a result of this finding, most fitness center instructors do not allow unr type squats.

Fry et al. (72) suggested that the r type squat could produce excessive forces at the hips and low back. Recently in a study of hip moments, (141; 142; 140) showed that r type squats result in larger hip moments compared with the unr type, whereas hip angles did not differ between the two squatting techniques. However, it is unclear what effect the restriction of displacement in the lower limbs has on the kinematics of the trunk. It can be assumed that compensatory motion of the trunk takes place when shank motion is restricted. Thus, it is necessary to investigate whole-body motion and especially trunk motion during squatting exercises. Thus, there is a need for objective assessment tools.

Most of the trunk models based on skin marker assessments that are currently in use either consider the trunk as a single segment (67; 121; 167; 237) or describe spinal motion (45; 71; 237). Crosbie et al. (44) divided
the trunk into three segments - the lumbar, lower trunk and upper trunk - each defined by three skin markers, allowing for the description of three-dimensional segmental kinematics. Locomotion and daily exercises were studied using a multi-segment trunk model (125). The authors used 14 markers for the pelvis and the trunk. Compared with the extensive range of different marker sets used to assess the kinematics of the lower extremities, very little work has been performed concerning the trunk.

The kinematic investigation of different types of squatting exercise is important for developing strategies for appropriate strength training (142). The aim of the present study was to compare lower extremity and trunk motion (dependent variables) between r and unr squats for the load conditions of 0% BW, 25% BW and 50% BW loading (independent variables). This included the development of a suitable trunk marker set and a kinematic procedure to assess the kinematics of the trunk during squats based on skin markers.

5.3 Methods

Experimental Approach to the Problem:

First, the subjects performed a standing trial in an anatomically upright position. To determine the ankle, knee and hip joint centers/axes of the skin marker model using functional approaches, four basic motion tasks (BMT) were performed (Table 1). Following these tasks, the subjects performed r and unr squats with zero, 25% bodyweight (BW) and 50% BW loading using a barbell. Each of the six conditions was performed for eight repetitions. In the r squat, the position of the most anterior part of the knee was not allowed to move beyond that of the toes. This restriction was visually self-controlled by the subject with the use of a live projection of the side view of knee and toes and a pile marking the front edge of the toes on a screen in front of the subject (140). With the latter set up no external force was
applied to restrict the motion of the shank and mimics a common training situation with guidelines of a coach and visual control in a mirror. The unr squat was performed with no restriction on the motion of the shank. The dependent variables were the joint angles and curvature of the spine and the independent variables were the execution type and the extra load.

**Subjects**

Thirty subjects, all movement science students experienced in weight lifting, with no history of back problems, participated in this study. The subjects had no specific sports activity background. The subjects were instructed to wear their normal sports shoes and to have a reasonable nutrition and hydration level. On average, the 30 subjects weighed 67 ± 11 kg, had heights of 174 ± 8 cm and were 25 ± 4 years. The study was approved by the Eidgenössische Technische Hochschule Zurich ethics committee. All subjects gave informed consent by signing the corresponding permission form.

**Procedures**

The 3D motion analysis system used was a 12-camera VICON MX system (Oxford Metrics Group, UK). The camera resolution was 2352 x 1728 pixels, the capture frequency used was 50 Hz and the capture volume was 300 cm x 500 cm x 200 cm. The instrumental error of marker position detection is ≤ 1 mm.

The marker set for the kinematic assessment (IfB Marker Set) consisted of 40 skin markers on the lower extremities, 7 markers on the pelvis and 24 markers on the trunk (Figure 1). The markers used had diameters of 9 and 14 mm. The allocation used of markers to segments is shown in Figure 1. Each segment is defined by a redundant marker cluster based on the following principles:

1. Marker visibility: All markers should be visible by at least two cameras
during the entire gait cycle.

2. Marker cluster distribution: The distance between markers and their offset from the lines joining the other markers should be as large as possible, thereby maximizing the mean cluster radius (207).

3. Number of markers: Each segment cluster consists of a redundant number of markers, as an increase from three to four or five markers improves the estimation of orientation accuracy (31).

4. Markers are located at positions on the body that show minimal skin movement artifacts.

The lower extremity marker set was used previously for the assessment of a level gait (134; 242), running (137; 242) and stair walking (106).

A segmental approach was applied to data from the whole body and the kinematics of the trunk was also assessed using a curvature approach. The data analysis was performed in Matlab R2010 (Natick, Massachusetts, USA). Segmental approach: The position and orientation of each segment was determined relative to the reference segments defined by the standing trial using a least-squares fit of the corresponding marker point clouds (74). It follows that the neutral position (0° rotation) was defined by the standing trial. Joint rotations were described from the distal relative to the proximal segment for the lower extremities (foot motion: forefoot relative to rearfoot, ankle motion: rearfoot relative to shank, knee motion: shank relative to thigh, hip motion: thigh relative to pelvis) and from the lower segment relative to the upper segment for the trunk (pelvic relative to lumbar, lumbar relative to thoracic segments). A helical axis approach was used (243). To define clinically interpretable rotational components, the attitude vector was decomposed along the axes of the marker-based joint coordinate systems (243).
Figure 1: The IfB Marker Set and joint coordinate systems used, including functionally estimated joint centers (yellow markers). The entire marker set consisted of 71 markers (because of a shortage of space, foot markers are only shown on the left leg of the skeleton). Red markers are used for segment tracking and to define the marker-based joint coordinate system; grey markers are only used for segment tracking. The hip joint center is used as a virtual marker for the thigh segment.
Table 1: Basic motion tasks (BMT) performed for the functional estimation of the ankle and hip joint centers and the knee joint axis (uk).

<table>
<thead>
<tr>
<th>Hip BMT</th>
<th>Three circumduction motions in the hip, standing on 1 leg, while keeping the leg straight. Support on the contralateral side was allowed.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee BMT</td>
<td>Loaded knee flexion/extension motions (3 repetitions). Bending the loaded knee while keeping the leg in the sagittal plane.</td>
</tr>
<tr>
<td>Ankle BMT1</td>
<td>Loaded inversion/eversion motion (3 repetitions). Movement of the shank in the frontal plane with a maximal range of motion relative to the standing foot.</td>
</tr>
<tr>
<td>Ankle BMT2</td>
<td>Loaded dorsiflexion/plantarflexion motion (3 repetitions). Movement of the shank in the sagittal plane with a maximal range of motion relative to the standing foot.</td>
</tr>
</tbody>
</table>

The ankle joint center (AJC), knee joint axis (uk) and hip joint center (HJC) were estimated via functional approaches. A functional approach was chosen to decrease the influence of anatomical landmark displacement and to allow higher accuracies than those obtained with prediction approaches (126). The ankle and hip joints were modeled for these purposes as ball-and-socket joints and the knee joint was modeled as a hinge joint. With the use of the corresponding BMT, as described in Table 1, the joint centers, respectively axes were determined by minimizing the sum of the differences between the modeled and the measured locations of the marker points. The marker-based joint coordinate systems are orthogonal right-handed coordinate systems built using the functionally estimated joint centers, respectively axes. The definitions for these coordinate systems are as follows and are illustrated in Figure 1:

- **Foot-joint coordinate system:** The posteroanterior axis eff connects HL and TO and is the leading axis. The direction of the mediolateral axis efs is perpendicular to eff and parallel to the ground. The vertical axis eff is perpendicular to the latter two axes.

- **Ankle-joint coordinate system:** The vertical axis eat, the connecting line between AJC and KJC, is the leading axis. The direction of the mediolateral axis eas is perpendicular to eat and lies in the plane spanned by the malleoli and the KJC. The posteroanterior axis eaf is perpendicular to the latter two axes.
• Knee-joint coordinate system: The mediolateral axis eks is defined by the functional estimated knee joint axis uk. The direction of the vertical axis ekt is perpendicular to eks and lies in the plane spanned by eks and the HJC. The posteroanterior axis ekf is perpendicular to the latter two axes.

• Hip-joint coordinate system: The mediolateral axis ehs is parallel to the line connecting the two AS markers. The direction of the vertical axis eht is perpendicular to ehs and lies in the plane spanned by KJC and ehs. The anteroposterior axis ehf is perpendicular to the latter two axes.

• Pelvic-joint coordinate system: The mediolateral axis eps is parallel to the line connecting the two AS markers. The direction of the vertical axis ept is perpendicular to eps and the plane spanned by the right and left AS and SACR.

• Trunk-joint coordinate system: The vertical axis ett, the connecting line between the spine markers L5 and C7, is the leading axis, and it points cranially. The transverse axis, ets, is perpendicular to ett and lies in the plane spanned by ett and the connecting line between the right and left BL markers, pointing from left to right. The posteroanterior axis etf is perpendicular to the latter two axes and points to the front.

Hence, clinical rotations are described as provided in Table 2.

Curvature approach for trunk kinematics: To assess the sagittal-plane curvature of the lumbar and the thoracic spine, corresponding marker positions were projected onto the sagittal trunk plane, defined as the plane spanned by ett and etf. The curvature was estimated by the reciprocal of the radius of a circle that was fitted by a least-squares approach into the five corresponding markers (Figure 2).
Figure 2: The IfB Marker Set and joint coordinate systems used, including functionally estimated joint centers (yellow markers). The entire marker set consisted of 71 markers (because of a shortage of space, foot markers are only shown on the left leg of the skeleton). Red markers are used for segment tracking and to define the marker-based joint coordinate system; grey markers are only used for segment tracking. The hip joint center is used as a virtual marker for the thigh segment.
Cycle definition

A squat cycle was defined as starting in a more or less upright position, moving down to the lowest position achieved during the squat and returning upward again. The start and end points of the cycle were defined by the vertical velocity of the barbell \( (v_{\text{barb}} > 0.02 \text{ m/s}) \) tracked by two markers attached to the ends of the barbell. For each condition, the mean and standard deviation (SD) were calculated over eight repeated cycles. Range of motion (ROM): The ROM was defined as the range between the minimal and maximal rotation values obtained.

Statistical Analyses

The influence of the squat technique used (r vs. unr) and the presence of an extra load on the joint angles and ROMs were analyzed using a multiple repeated-measures ANOVA. Eight valid executions were averaged for the statistical calculations of each technique. The significance was determined at \( p < 0.05 \). Statistical calculations were performed using IBM SPSS software version 19 (SPSS AG, Zurich, Switzerland).

5.4 Results

Segmental Trunk Motion

For all weight conditions and squat types, segmental trunk rotation was predominant in the sagittal plane, and rotation in the frontal and transverse planes was small (Figure 3). Comparing the unr technique to the r squat, the pelvis relative to lumbar segment flexion/extension, showed a small but not statistically significant increase in ROM in r over unr squats (Table 3). The ROM for flexion/extension between the lumbar and the thoracic segments was significantly larger for r squats compared with unr squats (Table 3). The ROM between the pelvis and the lumbar segments decreases
Figure 3: Trunk kinematics - segmental approach. Mean and SD of all 30 subjects are reported. Condition: restricted squat with 25% bodyweight loading.

significantly under the condition of a 50% BW load compared with 0% BW and 25% BW load conditions (Figure 4). The ROM between the lumbar and the thoracic segments increases significantly from 0% BW to 25% BW loads and significantly decreases from 25% BW to 50% BW loads. Thus, the largest ROM for the lumbar segment relative to the thoracic segment was observed for a 25% BW load (Figure 4).

Curvature of the trunk: The curvature of the lumbar and the thoracic spine decreases during the first half of the squat cycle, from upright standing to the lowest position of the cycle, and increases again during the second half of the cycle, during rising until the upright start position (Figure 5). When comparing the two squat techniques, the subjects show a significant increase in thoracic curvature ROM in the r squat condition compared with the unr condition. Equivalent behavior, but no significant difference, is observed in
Table 3: ROM of the segmental sagittal-plane rotations. Mean and SD of 30 subjects are reported.

<table>
<thead>
<tr>
<th></th>
<th>un00</th>
<th>un25</th>
<th>un50</th>
<th>r00</th>
<th>r25</th>
<th>r50</th>
</tr>
</thead>
<tbody>
<tr>
<td>Foot</td>
<td>2.0 ± 0.8</td>
<td>2.1 ± 1.0</td>
<td>2.4 ± 1.2</td>
<td>2.0 ± 1.0</td>
<td>1.9 ± 1.0</td>
<td>1.9 ± 1.0</td>
</tr>
<tr>
<td>Ankle*</td>
<td>29.6 ± 5.1</td>
<td>32.2 ± 4.6</td>
<td>32.8 ± 4.8</td>
<td>19.9 ± 3.8</td>
<td>21.9 ± 3.8</td>
<td>22.3 ± 3.3</td>
</tr>
<tr>
<td>Knee*</td>
<td>95.5 ± 9.8</td>
<td>100.7 ± 9.6</td>
<td>99.4 ± 8.1</td>
<td>76.4 ± 11.8</td>
<td>82.8 ± 11.5</td>
<td>84.0 ± 12.1</td>
</tr>
<tr>
<td>Hip</td>
<td>86.7 ± 10.1</td>
<td>83.1 ± 10.5</td>
<td>78.2 ± 10.2</td>
<td>85.3 ± 10.5</td>
<td>84.5 ± 12.4</td>
<td>82.3 ± 12.4</td>
</tr>
<tr>
<td>Pelvis rel lumbar</td>
<td>17.7 ± 5.9</td>
<td>17.0 ± 5.7</td>
<td>15.4 ± 4.6</td>
<td>18.6 ± 6.0</td>
<td>17.7 ± 6.5</td>
<td>16.3 ± 5.3</td>
</tr>
<tr>
<td>Lumbar rel thoracic segment*</td>
<td>7.9 ± 3.1</td>
<td>9.3 ± 3.3</td>
<td>8.7 ± 2.8</td>
<td>10.5 ± 4.7</td>
<td>12.1 ± 5.1</td>
<td>10.9 ± 4.6</td>
</tr>
</tbody>
</table>

*unr = unrestricted, r = restricted, 00 = 0% bodyweight loading, 25 = 25% bodyweight loading, 50 = 50% bodyweight loading.

Figure 4: Load dependency—mean and SD of all 30 subjects and both squat types (r and unr) are reported. * Significant difference between the loading conditions (p<0.05)
Figure 5: Trunk kinematics - curvature approach. Mean and SD of all 30 subjects are reported. Condition: restricted squat with 25% bodyweight loading.

Table 4: ROM of the segmental sagittal-plane curvature of the lumbar and the thoracic spine. Mean and SD of 27 subjects are reported.

<table>
<thead>
<tr>
<th>ROM of the curvature of the lumbar and thoracic spine</th>
<th>[1/m]</th>
<th>un00</th>
<th>un25</th>
<th>un50</th>
<th>r00</th>
<th>r25</th>
<th>r50</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lumbar</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>3.99 ± 1.59</td>
<td>2.65 ± 1.49</td>
<td>2.25 ± 1.31</td>
<td>3.58 ± 1.64</td>
<td>3.08 ± 1.66</td>
<td>2.32 ± 1.25</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Thoracic</td>
<td>0.03 ± 0.47</td>
<td>0.98 ± 0.54</td>
<td>0.95 ± 0.49</td>
<td>1.11 ± 0.52</td>
<td>1.24 ± 0.60</td>
<td>1.10 ± 0.58</td>
<td></td>
</tr>
</tbody>
</table>

The lumbar curvature under the two conditions (Table 4). The curvature of the spine shows a clear load dependency. There is a significant decrease in the lumbar curvature ROM with increasing loads (Figure 4). The thoracic curvature shows a significant decrease in ROM from 25% BW to 50% BW. However, from 0% BW to 25% BW, the thoracic curvature ROM slightly increases, but not significantly (Figure 4).

Segmental rotations of the lower extremities: Sagittal-plane rotations of the lower extremities were predominant, as observed in the trunk, and the transverse- and frontal-plane rotations were small (Figure 5). The observed
motion between the forefoot and the rearfoot was also marginal in the sagittal plane (Figure 6). The ROM of the knee and the ankle is significantly larger in unr type squats compared with r type squats (Table 3). The ROM of the foot and hip did not differ between the two techniques (r and unr) (Table 3). Under different loading conditions, the ROM of the ankle and the knee significantly increased, from 0% BW to 25% BW and 50% BW (Figure 4), respectively, whereas the ROM of the hip significantly decreased with an increase in load (Figure 4).

Cycle repeatability: The repeatability of waveforms within a test day is described by the coefficient of multiple correlation (CMC) (11). The average coefficient of multiple correlation of the vertical barbell motion (Figure 7) over all subjects and all six conditions was 0.99 (Table 5). Generally, sagittal-plane rotations exhibited larger CMCs than frontal- and transverse-plane rotations (Table 5). The CMC for sagittal-plane trunk rotation was larger for rotations between the pelvis and the lumbar segment than for rotations between the lumbar and the thoracic segments (Table 5).

5.5 Discussion

A marker set and a corresponding data processing protocol have been developed to assess the kinematics of the lower extremities and the trunk. The kinematics of the trunk were assessed using a 3D segmental approach using three trunk segments and one pelvic segment and by using a sagittal-plane spine curvature analysis. Both approaches are suitable for assessing the movement of the trunk during squatting.

Cycle repeatability was high for the barbell motion during squatting. Segmental rotations exhibited higher cycle repeatability for the sagittal plane than for the transverse and frontal planes. In the sagittal plane, the kinematics of the ankle, knee, hip and pelvis relative to the lumbar segment were highly repeatable between cycles, whereas the repeatability of
Figure 6: Lower leg kinematics—segmental approach. The mean and SD of all 30 subjects are reported. Condition: restricted squat with 25% bodyweight loading.
Figure 7: Vertical bar motion. Mean and SD of all 30 subjects are reported. Condition: restricted squat with 25% bodyweight loading.

Table 7: Coefficients of multiple correlation describing the repeatability of waveforms within a test day. The mean and SD of all subjects and all conditions are reported.

<table>
<thead>
<tr>
<th>[1/m]</th>
<th>rom00</th>
<th>rom25</th>
<th>rom50</th>
<th>ro00</th>
<th>ro25</th>
<th>ro50</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lumbar</td>
<td>3.39 ± 1.59</td>
<td>2.65 ± 1.49</td>
<td>2.25 ± 1.31</td>
<td>3.58 ± 1.64</td>
<td>3.08 ± 1.66</td>
<td>2.32 ± 1.25</td>
</tr>
<tr>
<td>Thoracic</td>
<td>0.93 ± 0.47</td>
<td>0.98 ± 0.54</td>
<td>0.95 ± 0.49</td>
<td>1.11 ± 0.52</td>
<td>1.24 ± 0.60</td>
<td>1.10 ± 0.58</td>
</tr>
</tbody>
</table>

*unr = unrestricted, r = restricted, 00 = 0% BW loading, 25 = 25% BW loading, 50 = 50% BW loading. Values are represented as mean ± SD. 

In comparison with unr, significantly different for all load conditions with p < 0.05.
the kinematics of the foot and lumbar relative to the thoracic segments and spine curvature was lower. It is unclear whether the cycle repeatability is primarily influenced by variance in execution by the subject or whether skin movement artefacts are also relevant influences. The cycle repeatability of joint rotations is lower than it is during level gait (112), as expected, because squatting is performed much less frequently than walking.

The smaller sagittal plane ROM occurring in the ankle joint in the r squat in comparison with the unr is given by the restriction demanded in the r technique and confirms that the squat has been executed correctly. The restriction in the motion of the shanks in the r type squat results in an increased ROM in the trunk, especially between the lumbar and the thoracic segments, but no increase in ROM in the hip occurs. The increased thoracic curvature in the r type squat represents increased flexion within the thoracic spine. Thus, a restriction of the shank results in an increase in trunk motion.

To prevent a backward fall, the restriction in shank motion needs to be compensated for by increased motion of the hips or the trunk to maintain stability with the center of mass vertically aligned above the base. The present data have revealed that this compensatory motion occurs in the trunk. Based on an assumption of a simple mechanical model, we can assume that a restriction in shank motion, resulting in increased trunk flexion, leads to higher stresses in the lower back. Extensive loading of the spine during r type squats has already been suggested by Fry et al. (72).

An increase in the extra load leads to an increase in the ROM of the ankle and knee and a decrease in the ROM of the hip and between the pelvis and the lumbar segment. It follows that with an additional load, the same range of movement of the barbell is reached using the distal joints rather than the proximal joints. An increase in the extra load leads to a smaller curvature of the lumbar spine, representing a straightening of the lumbar spine. This load-dependent straightening of the spine is in agreement with
the work of Meakin (156). The load conditions of the present study (0% BW, 25% BW, 50% BW) is a refinement of the loading range used in the study of McKean et al. (152). However, it has to be kept in mind that in a hypertrophy training the load may exceed this range. Although the present marker set included redundant marker point clouds to improve orientation accuracy (31), the work of Mörl and Blickhan (161) showed a close relation between a skin marker and its correspondent anatomical landmark on the vertebrae and that there is no impact during squatting, kinematic assessment is still limited by skin movement artefacts. The trunk coordinate system is based on markers and not functional approaches. Therefore, the influence of anatomical landmark displacement artefacts is a given, and testers should be well-trained in attaching markers. The last and potentially largest limitation of the present method in the assessment of trunk kinematics is that the segmental approach breaks the trunk into only three rigid segments. Future studies should compare skin marker trunk assessment methodologies with imaging methodologies that allow the direct assessment of the kinematics of the vertebrae by magnetic resonance imaging or videofluoroscopy.

5.6 Practical applications

In this study, whole-body kinematics was compared between two types of squatting over three loading conditions. For the trunk, the 3D segmental trunk motion was based on a three-segment trunk model, and the sagittal-plane spine curvature was determined. Generally, it was found that the lumbar spine exhibits load-dependent straightening. Not surprisingly, the ROM of trunk flexion during squatting increases with a restriction in shank motion. Therefore, we predict that there is less stress on the lower back during an unrestricted squat than during a restricted squat. Practitioners should not be overly strict with athletes/clients in coaching against anterior knee displacement during performance of the squat. A certain amount of
anterior displacement at the knee during unrestricted squatting may prevent undue stress on the lumbar spine and potential for back discomfort. Thus, when training leg muscles, the unrestricted squat may be the best choice.

Acknowledgements

This study was not funded. Alex Stacoff† was co-investigator in this project. We miss him.
In this study, the position of the spine during sitting was determined using open MRI and compared between different subjects. A wooden chair was used as a boundary condition during open MRI imaging. The motion in the spine between reclined upright and forward sitting posture involved all segments of the spine in all subjects. Non-uniform motion patterns were observed between subjects, especially during the forward sitting posture. This work was funded by Vitra AG.

**Contributions:** DB was the main investigator. SL was PI, designed the study and supported and supervised the data collection, evaluation and interpretation. All authors contributed to the text of the manuscript.
6 The spinal curvature of three different sitting positions analysed in an open MRI scanner

Daniel Baumgartner\textsuperscript{1,2}, Roland Zemp\textsuperscript{1}, Renate List\textsuperscript{1}, Mirjam Stoop\textsuperscript{1}, Jaroslav Naxera\textsuperscript{3}, Jean Pierre Elsig\textsuperscript{4} and Silvio Lorenzetti\textsuperscript{1}

\textsuperscript{1} Institute for Biomechanics, ETH Zurich, Switzerland
\textsuperscript{2} School of Engineering, Winterthur, Switzerland
\textsuperscript{3} Röntgeninstitut Zürich-Altstetten, Zurich, Switzerland
\textsuperscript{4} Spine Surgery, 8700 Küsnacht, Switzerland

Address correspondence to Silvio Lorenzetti, sl@ethz.ch

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6.1 Abstract

Sitting is the most frequently performed posture of everyday life. Biomechanical interactions with office chairs have therefore a long-term effect on our musculoskeletal system and ultimately on our health and wellbeing. This paper highlights the kinematic effect of office chairs on the spinal column and its single segments. Novel chair concepts with multiple degrees of freedom provide enhanced spinal mobility. The angular changes of the spinal column in the sagittal plane in three different sitting positions (forward inclined, reclined, and upright) for six healthy subjects (aged 23 to 45 years) were determined using an open magnetic resonance imaging (MRI) scanner. An MRI-compatible and commercially available office chair was adapted for use in the scanner. The midpoint coordinates of the vertebral bodies, the wedge angles of the intervertebral discs, and the lumbar lordotic angle were analysed. The mean lordotic angles were 16.7 ± 8.5° (mean ± standard deviation) in a forward inclined position, 24.7 ± 8.3° in an upright position, and 28.7 ± 8.1° in a reclined position. All segments from T10-T11 to L5-S1 were involved in movement during positional changes, whereas the range of motion in the lower lumbar segments was increased in comparison to the upper segments.

6.2 Introduction

During daily life, increasing amounts of time are spent in a sedentary position. In industrial countries, more than 75% of all office workers sit for periods of more than seven hours (132). In contrast to walking and running, muscles are not actively used during sitting. The muscular function is replaced by the supporting effect of the seat. Muscular inactivation over a long period of time leads to a weakening of the corresponding muscles. Approximately half of all office workers are affected by back problems (138), and recent trends show an increase in this number. Current research is therefore
focussed on sitting in relation to discomfort and pain (132). Grimmer et al. (81) found that adolescents have high rates of back pain which are medically verifiable and follow into adulthood. Accelerated degeneration of the spine due to long-term sitting results in a higher number of disc protrusions in the elderly (226). Further negative effects such as muscle clenching, nerve irritation, reduced blood circulation due to compressed veins, or narrowing of the respiratory organs may appear (226; 34; 211; 86). Such diseases may potentially cause chronic health problems in the elderly. A study by Katzmarzyk et al. (113) showed a higher incidence of cardiovascular disease and a higher risk of mortality for office workers compared with physically active working people, independent of their physical activity level during leisure time. These data were confirmed in a similar study performed by Patel et al. (176). A long-term sitting position therefore seems to be one of the highest risk factors for developing future health problems. This fact is also supported by a recent study of Dunstan et al. (50). They established that short bouts of walking during sitting time lower postprandial glucose and insulin levels in overweight/obese adults. Dunstan et al. (50) finally concluded that this may improve glucose metabolism and potentially be an important public health and clinical intervention strategy for reducing cardiovascular risk. Recent developments in the field of ergonomic office furniture allow different types of movements, such as forward and backward inclination as well as lateral tilting of the seat (82). A less constrained seat system leads to an alternating load on the spine, particularly on the intervertebral discs. Active but controlled sitting is believed to activate muscles and supporting structures and therefore prevent static loads acting on joints, ligaments, and tendons. It has been shown that an alternating sitting position significantly enhances muscular activity (55). Continuous upright sitting has been shown to be undesirable since the 1960s. Novel solutions with adjustable backrests or seats that alternate kyphosis and lordosis angles have been presented.
The kneeling chair represented one of the first sitting concepts that significantly influenced spinal posture. For example, Bennett et al. (16) found an increased lumbar curvature when sitting in a Balans Multi-Chair (kneeling chair) compared to sitting upright in a straight-backed chair. Other concepts included applying very small active rotational seat movements using motor-driven actuation, which resulted in a twisting of the spine along the vertical axis within the natural range of movement of individual intervertebral discs (130). These dynamic stimuli apparently influenced the length of the spine after sitting for a certain period of time. This continuous passive motion concept was previously published by Reinecke et al. (183) and was thoroughly investigated in later studies by Lengsfeld et al. (130) and van Deursen et al. (224). Recently, this actively steered chair was the focus of a biomechanical investigation by van Dieen et al. (225). The lordosis angle of different body postures and its effect on lumbar biomechanics have often been the focus of spinal research. Bridger et al. (21) concluded that the lordosis angle was smaller in a sitting position compared to a standing position. A forward tilted sitting position was therefore suggested in order to achieve similar lordosis angles as in standing. A reclined position reduces the load on the intervertebral discs and on the back muscles by an increased lordosis angle, which was shown by Colombini et al. (39). Graf et al. (77) demonstrated a certain discomfort with a tilting seat angle of more than 15°. Apparently, the biomechanical analysis of the lordosis angle is of relevance when determining the influence on posture and wellbeing. A more accurate analysis of the behaviour of single functional spinal units instead of the lordosis angle could be of advantage. Seat systems that allow several sitting positions have a significant influence on posture. The assessment of posture is hindered by the fact that the spine is positioned below soft tissue and the skin surface. The location of a single vertebra can only be assumed by the external shape of the thorax or by palpating the spinous processes
on the skin surface. Hence, a reproducible analysis method to quantify the location of single vertebrae is needed. Magnetic resonance imaging (MRI) techniques are therefore valuable for displaying the exact vertebral position in sedentary positions (17; 131).

By use of an upright, open MRI scanner, acquisition of 3-D data in the standing or sitting position is possible. In this way, the spine of wheelchair users has been investigated in a study by Linder-Ganz et al. (131). Savage et al. (193) and Videman et al. (226) correlated the clinical diagnosis displayed by MR images with the occurrence of symptomatic low back pain. Bertschinger et al. (17) compared sedentary patients in an open MRI scanner versus a traditional MRI scanner, in which patients have to lie down. In contrast to the standardised lying position in the closed-magnet unit, the spinal column is loaded by the gravitational weight of the thorax in the upright position. Thus, a sedentary position seems to be more clinically relevant in performing an accurate clinical diagnosis (56). Consequently, the analysis of variable sedentary positions on office chairs and the influence of these positions on spinal biomechanics can be accurately analysed with an open, upright MRI scanner. Dynamic or active sitting occurs when a chair enables the seated user to move in different planes. Flexibility and movement during sitting may be beneficial to wellbeing and allow different movement tasks to be performed. In our specific case, dynamic movement denotes a forward tilting mechanism of the seat pan. In particular, a higher degree of freedom for the hip flexural angle is provided which substantially influences the lumbopelvic mechanism and consequently the whole thoracolumbar region of the spine. The aim of this study is to analyse the spinal shape and, in particular, the position of single intervertebral bodies in relation to the sitting posture. The results may help to evaluate novel designs of backrests from a physiological point of view.
6.3 Material and Methods

Subjects

Six subjects (three females and three males, average age: 32 years (range 23 years to 45 years), average height: 1.74m (range 1.64m to 1.78 m), average weight: 68 kg (range 60 kg to 77 kg)) were measured in the three different positions. The subjects needed to have a maximum trunk width of 48 cm (distance from left to right shoulder) to have enough space in the MRI scanner. Clinical and therapeutic interventions relating to back problems were exclusion criteria. These criteria included previous back surgeries, diagnosed postural deformities in the sagittal or frontal plane, and presence of ferromagnetic implants in the body. Before measurements were made, metallic objects such as necklaces or watches were removed. No financial compensation was provided for participation in the study. A survey was filled out by every subject describing body parameters and history of back problems and clinical interventions. The study was approved by the ethics commission of the ETH Zurich (no. EK 2010-N-27).
Investigated Positions

The spinal posture and the position of the lumbar and lower thoracic vertebrae were analysed. Subjects were positioned on a specifically designed, MRI-compatible chair in the upright MRI scanner. The chair did not contain any ferromagnetic assemblies to exclude image artefacts. The duration of the scanning period depended on the size of the subject and was approximately three to five minutes per position. During measuring, the subject had to maintain a static position as far as possible. Three different chair positions were analysed (Figure 1). Upright (up). The lumbar spine was in contact with the backrest, but no force was transmitted. The hands were placed on the legs. Reclined (re). The back had contact with the whole backrest of the chair, the hands were placed on the legs, and the head was kept looking straight ahead. The subject was able to choose the most individually appropriate position. Forward Inclined (fi). The back had no contact with the backrest, and the upper body was supported by the arms lying on a table in front of the subject. A randomised sequence of the positions for every patient was performed. While changing from one position to another, a recovery time of five to ten minutes was given. During that time, the subjects were requested to walk around and relax the musculoskeletal system.

Data Acquisition and Measuring Sequence

Measurements were taken in the Upright MRI Center, Zurich, with the FONAR Upright MRI scanner (0.6 Tesla). T2-weighted sagittal images were taken with a repetition time of 3435 ms, an echo time of 110 ms, and a layer thickness of 4 mm. The resolution was 240 x 240 pixels in an image plane of 380 x 380 mm. In total, 15 sagittal sections were obtained for a vertebra with a 60mm width. The scans were captured along the vertical axis of the spine in sagittal sections. In a lateral view, the cross-section of the single
discs and their adjacent vertebrae are displayed. Discs and vertebrae can be easily separated due to the different contrasts, which is a result of the increased water content of the discs compared to the vertebral bone. The vertebral end plates between the discs and the vertebrae are displayed in a dark colour (Figure 2(a)).

**Data Analysis**

The following parameters were evaluated based on the MR images of the most central section (median plane of the body).

*Coordinates of the Midpoints of the Vertebrae* A coordinate system was placed corresponding to the main direction of the MR image with the origin through the lowest vertebra L5 (Figure 2(a)). The x- and y-coordinates were determined for all three sitting positions in the sagittal view for each vertebra (from L5 to Th10). The coordinates were determined based on the centre of a quadrangle built by the two endplates and the ventral and dorsal margins of the vertebral bodies.

*Wedge Angles of the Intervertebral Discs* The wedge angles (from L5/S1 to Th10/Th11) were determined according to an established, clinical evaluation (157). A tangent line was placed on the ventral and dorsal edges of each vertebra. The wedge angle was defined in between two lines of adjacent vertebral bodies (Figure 2(b)).

*Lordotic Angle* The angle between the tangent line on the upper L1 endplate and the tangent line on the upper sacrum S1 endplate is defined as the lordotic angle $\alpha$ (Figure 2(b)).

*Convention* A lordosis was defined as a positive angle and a kyphosis as a negative angle.

MegaCAD 2D software (Version 2011, MegaCAD-Center GmbH, Oberweningen, Switzerland) was used to examine the coordinates and angles.
Figure 2: (a) The coordinate system (red arrows) and the quadrangle built by the two endplates and the ventral and dorsal margins of the vertebral body (yellow straight lines) to determine the coordinates of the vertebral midpoints (red stars). (b) The wedge angles of the intervertebral discs ($W_{L5/S1}, W_{L4/L5}, \ldots$) and the lordotic angle $\alpha$. 
**Statistical Analysis**

All statistics were determined using IBM SPSS Statistics (Version 19, SPSS Inc., Chicago, IL, USA). The statistical significance level was set at $P < 0.05$.

The averages of the wedge angles and the lordotic angle of the three positions (up, re, fi) were compared with the Wilcoxon tests in a crosswise manner.

### 6.4 Results

**Coordinates of the Midpoints of the Vertebrae**

Changes in the position and shape of the spine occurred during the three different sitting positions (Figure 3). The reclined sitting posture resulted in similar positions of the vertebrae for all subjects. In contrast, the largest differences in the position of the midpoints between the subjects occurred when patients were in the forward inclined sitting posture. In the upright sitting posture, five subjects had a slightly dorsally located spine. Only one subject (Subject 3) showed a ventral configuration of the vertebrae, especially in the lumbar region. For all subjects, the shape of the lumbar spine was similar during the upright and the inclined positions. A line through the midpoints of the vertebrae, approximating the direction of the spine in the reclined position, was dorsally ascending with an angle between 30° and 40° relative to the vertical axis. The upright position was dorsally ascending with a large variability for all subjects.

**Wedge Angles**

The maximal measured mean wedge angle was $3.4 \pm 1.2^\circ$ (mean ± standard deviation) for the lowest lumbar segment in the reclined position. The maximal
Figure 3: Coordinates of the midpoints for the vertebrae of the three positions. All curves are related to the same origin, represented by the midpoint of L5.
mean changes within one segment were $3 \pm 2^\circ$ (mean $\pm$ standard deviation) from the forward to the reclined position for the wedge angle TH12/L1. Generally, a change in position was visible for all segmental heights, and some were significant (Figure 4). A general trend of a uniform movement pattern was not observed for the six subjects. Two subjects reached their maximum wedge angle in the upright position, while all others reached this angle in the reclined position.

Figure 4: Mean wedge angles and their standard error of the intervertebral discs. *Significant differences between positions.
Figure 5: Mean lordotic angles ($\alpha$) and their standard deviation for the three positions.

**Lordotic Angle**

The mean lordotic angle $\alpha$ for the forward inclination was $16.0 \pm 8.5^\circ$(mean ± standard deviation), for the upright position was $24.7 \pm 8.3^\circ$, and for the reclined position was $28.7 \pm 8.1^\circ$(Figure 5). High individual differences were observed for the lordotic angle. These large interindividual differences were also observed for the reclined position, although a standardised backrest was used. The lordotic angles were not significantly different between the three positions.

**6.5 Discussion**

All lumbar and lower thoracic intervertebral discs are involved in positional changes and contribute to the change in the spinal shape. These findings were revealed by applying the clinical evaluation method to determine the wedge angles. No specific segment can be identified in which the majority of the movement is performed. Slight trends in the absolute wedge angles of the
intervertebral discs could be positions seem to be higher for the lower lumbar vertebral discs and decrease towards the tenth thoracic vertebra. In contrast, individual differences between the subjects were much higher for the upright position compared with the other positions. No general movement pattern caused by changing positions was detected. High individual differences are visible, although the geometry of the test chair was standardised for all tested subjects. The current study was performed with only 6 subjects, which represents the main limitation. Only some of the analysed spinal angles were significantly different. To provide more statistically significant data, more subjects would be required. However, some general statements about the behaviour of the vertebral discs of the lower back could be made. In conclusion, the wedge angles and the position of the vertebral bodies change between the three described sitting positions. As a result, the load condition of the intervertebral discs changes. This is assumed to stimulate the metabolism of the intervertebral discs (155). Even slight changes in the position cause a change in the disc loading. Positional changes from an upright to a reclined or forward inclined sitting position may therefore have a positive effect on the biological nutrition processes of the spine.

**Funding**

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**Acknowledgments**

The authors would like to thank Nadja Mettler for the technical support at the Upright MRI Center Zurich.
It is important to assess the accuracy of a measurement tool before using it for scientific research. In this study, the accuracy of a novel skin marker set in measuring the curvature of the spine and segmental angles of the trunk was assessed using open MRI. Large errors in the absolute values of the skin marker positions were found compared to the positions of the underlying bone segments based on image data from open MRI. However, changes in the positions of the skin markers, i.e., changes in the shape of the spine due to motion, could be measured with reasonable accuracy using the proposed marker set.

**Contributions:** This work is based on the master thesis of RZ. Conceived and designed the experiments: RZ SL. Performed the experiments: RZ JPE JN. Analyzed the data: RZ RL TG JN JPE WRT SL. Contributed reagents/materials/analysis tools: JN JPE. Wrote the paper: RZ RL TG WRT SL.
Soft tissue artefacts of the human Back: Comparison of the sagittal curvature of the spine measured using skin markers and an open upright MRI

Roland Zemp¹, Renate List¹, Tugut Gülay¹, Jean Pierre Elsig², Jaroslav Naxera³, William R. Taylor¹ and Silvio Lorenzetti¹

¹ Institute for Biomechanics, ETH Zurich, Switzerland
² Spine Surgery, 8700 Küsnacht, Switzerland
³ Röntgeninstitut Zürich-Altstetten, Zurich, Switzerland

Address correspondence to Silvio Lorenzetti, sl@ethz.ch

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7.1 Abstract
Soft tissue artefact affects the determination of skeletal kinematics. Thus, it is important to know the accuracy and limitations of kinematic parameters determined and modelled based on skin marker data. Here, the curvature angles, as well as the rotations of the lumbar and thoracic segments, of seven healthy subjects were determined in the sagittal plane using a skin marker set and compared to measurements taken in an open upright MRI scanner in order to understand the influence of soft tissue artefact at the back. The mean STA in the flexed compared to the extended positions were 10.2±6.1 mm (lumbar)/9.3±4.2 mm (thoracic) and 10.7±4.8 mm (lumbar)/9.2±4.9 mm (thoracic) respectively. A linear regression of the lumbar and thoracic curvatures between the marker-based measurements and MRI-based measurements resulted in coefficients of determination, $R^2$, of 0.552 and 0.385 respectively. Skin marker measurements therefore allow for the assessment of changes in the lumbar and thoracic curvature angles, but the absolute values suffer from uncertainty. Nevertheless, this marker set appears to be suitable for quantifying lumbar and thoracic spinal changes between quasi-static whole body postural changes.

7.2 Introduction
Back pain is an increasingly common affliction, with approximately one-third of the population suffering from low back pain at any given time (148). Kinematic parameters of the lumbar spine, such as the rate of angular rotation and linear displacement at the joints (L3/L4; L4/L5; L5/S1), especially during the onset of lumbar flexion, are useful for discriminating between individuals with and without low back pain (219). While motion of the lumbar spine is accessible using video fluoroscopy (219; 128; 220), the approach is highly invasive and exposes subjects to unnecessary X-ray radiation (220). Moreover, while novel dynamic and non-invasive approaches now exist for
assessing functional motion of the back, even over extended periods of time (40; 41; 42; 216), the accuracy of such methods for evaluating the underlying skeletal kinematics remains unknown.

In addition to instability and degeneration of supporting soft tissue structures, overloading is considered a main cause of low back pain due to a combination of cumulative or acute loads (123; 102). However, determination of the internal loading conditions requires knowledge of the spine’s position and movement. While bone pins allow direct access to skeletal kinematics (188; 120; 12), they are only rarely used due to their invasive nature. Motion analysis, on the other hand, allows the non-invasive investigation of motion patterns (218). However, while skin markers are easy to apply and rarely limit the subject’s movement, they are affected by soft tissue artefact (STA), which results from motion of the skin relative to the underlying bones due to inertial effects, skin elasticity and deformation caused by muscle contraction (28; 217; 127). STA occurs in all directions, and the distributions are known to be non-uniform (75).

To obtain an understanding of the accuracy of skin markers for assessing spinal kinematics, a marker set has been developed that allows global parameters of curvature angles (129; 114; 241; 3; 172; 8; 71) and back segment rotations (24; 247; 25; 215; 43) to be investigated (135). However, the accuracy and precision of these analyses remain unknown. Several studies have validated skin markers on different body regions (75), (190; 147; 221; 146; 76; 189; 2), but validation studies of back markers are rare (24; 247; 25; 161; 94) (Table 1), and few have been validated against global spinal shape, including spinal segment curvature or rotations.
7.3 Methods

Accuracy of the marker and vertebra position with MRI

The accuracy of assessing vertebral location using MRI was determined using a plate of acrylic glass with five MRI-visible skin markers (paintballs) and two lamb vertebrae (Figure 1). The paintballs were placed into precision-drilled holes (±0.1 mm) in each of the four corners and middle of the plate. The two vertebrae were glued onto the plate between the markers. The plate was then examined using an open MRI (Upright Multi-Position MRI; 0.6 Tesla; Fonar Corporation, Melville, USA) in horizontal (0°), forward tilted (45°) and vertical (90°) positions in order to quantify the accuracy of the marker and vertebrae locations as well as the orientation of each vertebrae base plate plane (BPP) relative to horizontal.

Subjects

Seven healthy subjects (three female; average age 29 y (range 22–46); height 174 cm (160–184); mass 71 kg (55–96)) provided written informed consent
to participate in this pilot study that was approved by the local ethics committee. Subject recruitment was achieved through voluntary participation after public poster advertising. The participant on Figure 2 has seen this manuscript and figure and has provided written informed consent for its use in publication. A wide range of subject height and weight was chosen in order to exemplarily investigate the range of kinematics that could be observed within a broad population. A power analysis (one-tailed paired t-test, $\alpha = 0.05$, $\beta = 0.1$) performed using a statistical software package (G*Power 3.1.3) (66), based on the determined accuracy of the acrylic plate measuring system (SD of the curvature angle due to the measurement system: $3.6^\circ$) and one test measurement (difference in lumbar curvature angle between the upright and extended position: $8^\circ$) with an effect size $dz$ of 1.571 (where $dz = 8^\circ/(3.6^\circ\cdot\sqrt{2})$) revealed a minimum subject number of six with a power level of 0.948.
Figure 2: Measurement set-up including (a) the "IfB-marker-set" of the trunk and the pelvis (for explanation of abbreviations and for segmental allocation, see Table 2) and (b) the three analysed seating positions with an example of a corresponding MR image including a local coordinate system of a vertebral body. Upright seating position: lower spine had partial contact with the backrest, and the whole upper body was in an upright position. Flexed seating position: upper body was tilted about 30° forward and supported on a bar, while the arms were rested on their lap. Extended seating position: the subject’s bottom was pushed approximately 20 cm forward and the head was supported by the backrest. (from left to right).
Instrumentation

T2-weighted sagittal images were taken with a repetition time of 2750 ms, an echo time of 110 ms and a layer thickness of 4 mm. The resolution was 240×240 in an image plane of 360×360 mm, providing a voxel size of 1.5×1.5×4 mm³. The layers were ranked without any gaps between the marginal markers. As a consequence, the lumbar and thoracic regions required approximately 35 and 50 images respectively, corresponding to a measurement time of approximately seven minutes for each posture.

Marker set

Based on our whole body "IfB-marker-set" (46; 133; 223; 210; 85; 140) (Figure 2a), only the markers for the lumbar and thoracic segments were used (Table 2). The markers were MRI-visible commercial paintballs (BrassEagle Wild Streak Paintballs, diameter 17.3 mm), which consisted of a dyed liquid surrounded by a thin gelatin shell. After palpation, performed in an upright standing position, the markers were mounted on washers and were fixed to the skin using a toupee plaster. In order to provide support during sitting while preventing marker contact with the backrest, two foam tubes were attached to the paraspinal muscle bellies. The subjects' lumbar and thoracic spines were then measured in the MRI in upright, flexed and extended seating positions (Figure 2b).

Data analysis

The MR images were manually segmented using Avizo (v5.1, Mercury Computer Systems Inc., Burlington, USA). Spheres were fitted to the markers (Geomagic Studio, v9, Raindrop Geomagic, USA), and the normal vector of each vertebral body's BPP and centre of gravity (CoG) were determined.

Data analysis was performed using MATLAB (vR2010a, MathWorks Inc., Natick, USA). STA was described by changes in the vectors pointing
from the vertebral bodies’ CoG to the corresponding marker. The normal vector of the BPP defined the cranial (z) axis of each vertebral body coordinate system. As an exception, due to increased segmentation stability, the z-axis of the fifth lumbar vertebra was determined using the upper plate, and rotated accordingly. The y-axis was the cross-product of the z-vector with the unit vector in the anterior-posterior direction of the MRI, and the x-axis was defined by the normalised cross-product of the y and z vectors (Figure 2b). Each local coordinate system was located at the CoG of the respective vertebral body. The vectors from the vertebral bodies’ CoG to the corresponding markers were constructed using the local coordinate systems. The differences between this vector in the upright and the flexed or extended seating positions described the magnitude and direction of STA of each marker on the spinous process.

To determine the curvature angles [48] of the lumbar (α_lumbar) and thoracic (α_thoracic) spines, the sagittal planes of the spines were defined normal to the vectors from RTBL to LTBL (V1) and from RTSC to LTSC (V2) respectively (Figure 2a). The position vectors of the markers and the vertebral bodies were projected onto this plane. Circles [49] were created for

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**Table 2: Marker placement, segmental allocation and abbreviations.**

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<thead>
<tr>
<th>Abbreviations</th>
<th>Marker placement</th>
<th>Segment allocation</th>
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<tr>
<td>RTSH, LTSH</td>
<td>Right and left axillary</td>
<td>Upper trunk segment</td>
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<tr>
<td>RTCL, LTCL</td>
<td>Right and left clavicle</td>
<td></td>
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<tr>
<td>C7</td>
<td>7th cervical vertebra</td>
<td></td>
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<tr>
<td>Cl, Cs</td>
<td>3rd, 5th cervical vertebra</td>
<td></td>
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<tr>
<td>STER</td>
<td>Sternum</td>
<td>Thoracic segment</td>
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<td>RTSC, LTSC</td>
<td>Right and left inferior angle of the scapula</td>
<td></td>
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<tr>
<td>RTBL, LTBL</td>
<td>Right and left most inferior rib</td>
<td>Lumbar Segment</td>
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<tr>
<td>L3, L5, T5, T11</td>
<td>3rd, 5th, 7th, 11th thoracic vertebrae</td>
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<tr>
<td>RTBL, Ltbl</td>
<td>Right and left lateral back on height of L4</td>
<td></td>
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<tr>
<td>L1, L2, L3, L4, L5</td>
<td>1st, 2nd, 3rd, 4th, 5th lumbar vertebrae</td>
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<td>RTPS, LTPS</td>
<td>Right and left anterior superior iliac spine</td>
<td>Pelvic segment</td>
</tr>
<tr>
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the lumbar and thoracic spines that best fitted the CoGs of L1–L5, and T3, T5, T7, T9 & T11 respectively. $\alpha_{\text{lumbar}}$ was then calculated as the angle between the two radius vectors from the circle centre to the CoG of L1 and L5, and $\alpha_{\text{thoracic}}$ accordingly as the angle between the radius vectors T3 & T11. The same angles were calculated for the corresponding lumbar and thoracic markers. Kyphosis was defined as a positive angle ($\alpha > 0$).

To analyse the accuracy with which the skin markers were able to represent the rotation of the vertebral bodies, the mean sagittal rotation error (ESR) of the lumbar and thoracic segments (Table 2) was calculated. Marker cloud registration was performed using a least squares method. The sagittal rotation of the lumbar and thoracic segments was calculated between the corresponding marker cloud in the upright and compared to the flexed or extended positions. Here, the rotation of each vertebral section was calculated using a 3D regression line, fitted through the vertebral CoGs, and compared against the rigid rotation of the relevant marker cloud.

Due to the large radius of the paintballs, some lumbar markers of subjects 3, 4, and 7 touched each other in the extended seating position and it was not possible to analyse these MR images. Owing to image blur as a result of body movement during the measurements, the thoracic MR images of subject 2 in the flexed seating position were not taken into account for the analysis.

**Statistics**

All statistics were determined using IBM SPSS Statistics (v19, SPSS Inc., Chicago, USA). Statistical significance was defined as $p < 0.05$. The absolute marker artefact was analysed using analyses of variance (ANOVA) for the subjects, positions and marker locations. Furthermore, correlations between the curvature angles based on the marker and vertebral body coordinates were investigated using linear regression analysis.
7.4 Results

Accuracy of the marker and vertebra positions using MRI

The accuracy of the MRI and the image segmentation was similar to that of a conventional motion capture system. Based on the mean error between different markers on the acrylic plate (0.6±0.5 mm) and between the vertebral CoGs and the markers (1.1±1.1 mm), the direction-related measurement uncertainties ($\sigma_x, \sigma_y, \sigma_z$) of the markers were (1.0 mm, 0.5 mm, 0.5 mm) and for the vertebral bodies (2.0 mm, 1.0 mm, 1.0 mm). The orientation of the BPP relative to horizontal varied by up to 2.7°. The mean error was 1.6±1.2°.

Subject measurements

The mean STA in the flexed and extended positions were 10.2±6.1 mm (lumbar)/9.3±4.2 mm (thoracic) and 10.7±4.8 mm/9.2±4.9 mm respectively. The largest STA was 27.4 mm for marker SPL5 in the flexed position. The STA was significantly different between subjects ($p<0.001$ lumbar and thoracic), but no differences were observed for either markers ($p = 0.604$ lumbar, $p = 0.404$ thoracic) or seating positions ($p = 0.428$ lumbar, $p = 0.926$ thoracic) (Table 3). The subject’s mean STA of the lumbar and thoracic markers as well as the flexed and extended positions varied between 6.2 mm and 13.2 mm for the seven subjects with a BMI between 20.6 kg/m² and 30.3 kg/m². However, no clear relationship between STA and BMI was observed.

The lumbar ($\alpha_{lumbar}$) and thoracic curvature angles ($\alpha_{thoracic}$) calculated using the markers and the vertebral bodies revealed no clear correlation ($R^2 = 0.552$ (lumbar); $R^2 = 0.385$ (thoracic); Figure 3). The root mean square errors (RMSEs) of the differences between the curvature angles determined by the markers and by the vertebral bodies were approximately two times
Table 3: Direction-related ($r_x$, $r_y$, $r_z$) mean marker artefact (mean) and the absolute values ($||r||$) with their standard deviations (SD) of the lumbar and thoracic skin markers in the flexed and extended positions.

<table>
<thead>
<tr>
<th>Marker</th>
<th>Flexion (mm)</th>
<th>Extension (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$r_x$ (mean SD)</td>
<td>$r_y$ (mean SD)</td>
</tr>
<tr>
<td></td>
<td>$r_x$ (mean SD)</td>
<td>$r_y$ (mean SD)</td>
</tr>
<tr>
<td>Thoracic</td>
<td></td>
<td></td>
</tr>
<tr>
<td>SPT1 (T3)</td>
<td>0.4 (1.8)</td>
<td>0.3 (1.3)</td>
</tr>
<tr>
<td>SPT2 (T5)</td>
<td>0.0 (4.4)</td>
<td>0.1 (2.3)</td>
</tr>
<tr>
<td>SPT3 (T7)</td>
<td>0.4 (3.7)</td>
<td>1.4 (2.3)</td>
</tr>
<tr>
<td>SPT4 (T9)</td>
<td>2.9 (4.5)</td>
<td>3.2 (4.2)</td>
</tr>
<tr>
<td>SPT5 (T11)</td>
<td>0.1 (4.8)</td>
<td>1.7 (3.6)</td>
</tr>
<tr>
<td>Lumbar</td>
<td></td>
<td></td>
</tr>
<tr>
<td>SPL1 (L1)</td>
<td>-1.8 (2.7)</td>
<td>-1.7 (2.7)</td>
</tr>
<tr>
<td>SPL2 (L3)</td>
<td>-2.6 (4.6)</td>
<td>-1.8 (2.2)</td>
</tr>
<tr>
<td>SPL3 (L5)</td>
<td>-3.7 (4.0)</td>
<td>-1.7 (2.6)</td>
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<td>SPL4 (L7)</td>
<td>-3.9 (4.8)</td>
<td>-2.6 (2.7)</td>
</tr>
<tr>
<td>SPL5 (L9)</td>
<td>-4.1 (3.6)</td>
<td>-2.6 (3.6)</td>
</tr>
</tbody>
</table>

doi:10.1371/journal.pone.0095425.t003
The x-axis represents the values calculated from the markers, and the y-axis from the vertebral bodies. The crosses show the open MRI’s measurement uncertainty of the curvature angles calculated from the markers (x-axis) and calculated from the vertebral bodies (y-axis).

higher for the lumbar spine than the thoracic spine (Table 4). The lumbar curvature angles from the upright to the flexed and extended positions showed the same sign in six of the seven subjects and three of the four subjects respectively, whereas the sign of the thoracic curvature angle was the same in all subjects (Figure 4).

The ESR of the lumbar and thoracic segments calculated using the skin markers were $2.5 \pm 2.7^\circ$ (RMSE: 3.6°) and $-1.1 \pm 2.9^\circ$ (RMSE: 3.0°) respectively. The largest ESRs were 6.6° (lumbar) and 9.1° (thoracic).

7.5 Discussion

While the exact motion of the vertebrae remains unclear without the use of invasive approaches, knowledge on the accuracy of skin markers for assessing skeletal kinematics provides a baseline for identifying situations where non-
Figure 4: Range of lumbar (a/b) and thoracic (c/d) curvature angle of the subjects, calculated using the vertebral bodies (black) and the skin markers (hatched). The range was defined from the upright to the flexed (a/c) and to the extended positions (c/d).

Table 4: Mean (SD) lumbar ($\alpha_{lumbar}$) and thoracic curvature angle ($\alpha_{thoracic}$) in upright, flexed and extended sitting position calculated with the vertebral bodies and the skin marker, as well as the mean differences (SD;RMSE) between the values from the skin marker and the vertebral bodies.

<table>
<thead>
<tr>
<th></th>
<th>Upright</th>
<th>Flexion</th>
<th>Extension</th>
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<tr>
<td></td>
<td>Vertebral bodies</td>
<td>Marker</td>
<td>Vertebral bodies</td>
</tr>
<tr>
<td>Difference ($\alpha_{lumbar}$)</td>
<td>(-16.8 (11.5))</td>
<td>(0.3 (6.3))</td>
<td>(-7.9 (10.2))</td>
</tr>
<tr>
<td></td>
<td>17.1 (8.0, 18.6)</td>
<td>17.3 (11.2, 20.2)</td>
<td></td>
</tr>
<tr>
<td>Difference ($\alpha_{thoracic}$)</td>
<td>36.0 (10.4)</td>
<td>46.7 (10.3)</td>
<td>36.4 (10.2)</td>
</tr>
<tr>
<td></td>
<td>4.7 (1.4, 11.5)</td>
<td>4.0 (9.4, 9.5)</td>
<td>4.0 (9.4, 9.5)</td>
</tr>
</tbody>
</table>
invasive approaches are appropriate, and where not (136). The measurement uncertainty of our MRI-based measurement system was similar to those of a typical motion capture system (1.5 mm) (10), with out-of-plane error about double the in-plane error. Analysis of the back STA produced similar results to those observed in other studies (161; 94) and also for other body parts (Table 1). The observed intra- and inter-individual patterns of STA during flexion and extension did not allow the determination of a common correction method by which to estimate the behaviour of single markers. In general, the inter-individual differences were larger than the differences within a single subject - a result that is in agreement with results from studies on the knee joint (76). Knowledge of the influence of STA when using skin markers is required to ensure compensation for the largest error sources (217; 122; 93).

The present study only allowed the quantification of STA in a static set-up. However, we must be conscious of the fact that the spinal movement differs from other skeletal joints e.g. the hip or the knee, since the spine consists of several segments that allow movement relative to one another, including a different range of motion (ROM) in several planes. The present study only allowed the quantification of STA in a static set-up. It must be assumed that during dynamic activities, and especially impact situations, STA is even larger. For example, Akbarshahi and co-workers [38] found much larger marker STA during functional activities than Sangeux et al. (190), who used a static set-up (Table 1). Therefore, studies using static measurements seem to underestimate the magnitude of STA, possibly by a factor of two or more.

Due to the fact that the power analysis in this study was based on measurements using lamb vertebrae, the actual study in humans may require additional subjects to ensure sufficient power, due to secondary errors such as unintentional movements during MRI scanning. However, as a result of
this power analysis, relatively few subjects were recruited into this study, and it was not possible to observe any clear relationships between STA and e.g. age, gender, height or material properties of the soft tissues. It is, of course, entirely possible that these or other subject-specific factors contribute to the magnitude of STA on the back. Here, it is quite conceivable for example that the individual elastic properties of the soft tissues have a strong influence on the magnitude of STA; such observations have been reported at other regions of the body (thigh) by Kratzenstein and co-workers (122), who also demonstrated locally varying STA, which was mainly attributed to muscle contraction and skin elasticity. In addition, no relationship between the levels of STA and BMI were observed in our study, but this could also be an artefact of the low number of subjects in our cohort. However, this finding is consistent with studies that used fluoroscopy (76) or bone pins (100) to examine the role of soft tissues on the underlying skeletal kinematics. Increasing the number of subjects could allow a better understanding of the relationships between BMI, age, gender, height or material properties of the soft tissues, but this was not the focus of the current study. In order to establish the influence of such subject-specific factors on STA, further research would be required in specific homogenous cohorts.

Since the correlations between the spinal curvature calculated from the skin markers and the vertebral bodies for each position (upright, flexion, extension) as well as for all positions together were low (Figure 3: $R^2 = 0.552$ (lumbar), $R^2 = 0.385$ (thoracic)), results that examine the lumbar and thoracic curvature angles by means of skin markers should be interpreted cautiously. This was possibly due to the fact that the anatomical distance and the material properties of the musculoskeletal tissues between the markers and CoG are generally not constant, resulting in inhomogenous deformation between the different positions. However, the range of curvature angles exhibited the same sign when comparing the upright with the
flexed and the extended seating positions in 22 out of the 24 cases (Figure 4). Due to the fact that the range of curvature calculated from the skin markers did not consistently over- or underestimate that calculated from the vertebral bodies within subjects, positions or spinal segment, there is no clear method to enhance the accuracy of skin marker estimations through automated correction.

To summarise, the results of our study indicate that a change of lordotic/kyphotic shape, but not of the absolute amount of curvature, can be estimated using skin markers. Based on these findings, the use of the presented back marker set for analysing spinal motion seems to be as accurate as estimations of skeletal kinematics in the lower extremities (Table 1). Changes of the lumbar and thoracic curvature angle are measurable in the sagittal plane using the presented marker set, but measurement of the absolute curvature angles appears to be limited when using skin markers. These limitations associated with STA must be taken into account during non-invasive assessment of back motion before an improved understanding of the kinematics of subjects with and without back pain can be gained.

Acknowledgments

We would like to thank the technical support of Kerstin Wenker at the Upright MRI Center Zürich. Ethics: This study was approved by the ethics committee of the ETH Zürich under the number: EK 2010-N-43.
In this paper, the motion and the loading state of the back during two common strength exercises, namely, deadlifts and good-mornings, were determined using data from optical motion capture and inverse dynamics analysis. A newly developed and validated marker set was used to capture the motion of the spine during exercise execution. The results from the present work will allow therapists and coaches to rate different types of strength exercises based on biomechanical principles and thus assist in the choice of suitable exercises to minimize injury.

The findings of the present work related to squatting were published in a popular non-scientific journal to promote the results amongst strength training practitioners (F. Schellenberg and S. Lorenzetti. 2014, Bewegung und Belastung bei den Goodmornings und dem Kreuzheben, *Fitness Tribune*, 153).

**Contributions:** This work is part of the PhD of FS. FS and JL acquired the data, and undertook the data analyses including the preparation of the tables and figures. FS furthermore performed the statistical analysis and helped drafting the manuscript. RL supported the measurement set up, data analysis and helped drafting the manuscript. BT helped in both interpreting the data and drafting the manuscript. SL made the concept and design of this study, supervised the data analyses and interpretation and helped drafting the manuscript. All authors read and approved the final manuscript.
8 Kinetic and kinematic differences between dead-lifts and goodmornings

Florian Schellenberg 1, Julia Lindorfer 1,2, Renate List1, William R. Taylor1 and Silvio Lorenzetti1

1Institute for Biomechanics, ETH Zurich, Switzerland
2Technikum Wien, Höchstädtplatz 6, 1200 Wien, Austria

Address correspondence to Silvio Lorenzetti, sl@ethz.ch

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8.1 Abstract

**Background:** In order to improve training performance, as well as avoid overloading during prevention and rehabilitation exercises in patients, the aim of this study was to understand the biomechanical differences in the knee, hip and the back between the exercises "Goodmornings" (GMs) and "Deadlifts" (DLs).

**Methods:** The kinetics and kinematics of 13 subjects, performing GMs and DLs with an additional 25% (GMs), 25% and 50% (DLs) body weight (BW) on the barbell were analysed. Using the kinetic and kinematic data captured using a 3D motion analysis and force plates, an inverse approach with a quasi-static solution was used to calculate the sagittal moments and angles in the knee, hip and the trunk. The maximum moments and joint angles were statistically tested using ANOVA with a Bonferroni adjustment.

**Results:** The observed maximal flexion angle of the knee was 5.3 ± 6.7° for GMs and 107.8 ± 22.4° and 103.4 ± 22.6° for DLs with 25% and 50% BW respectively. Of the hip, the maximal flexion angle was 25% smaller during GMs compared to DLs. No difference in kinematics of the trunk between the two exercises was observed. For DLs, the resulting sagittal moment in the knee was an external flexion moment, whereas during GMs an external extension moment was present. Importantly, no larger sagittal knee joint moments were observed when using a heavier weight on the barbell during DLs, but higher sagittal moments were found at the hip and L4/L5. Compared to GMs, DLs produced a lower sagittal moment at the hip using 25% BW while generating the same sagittal moment at L4/L5.

**Conclusions:** The two exercises exhibited different motion patterns for the lower extremities but not for the trunk. To strengthen the hip while including a large range of motion, DLs using 50% BW should be chosen. Due to their ability to avoid knee flexion or a knee flexion moment, GMs should be preferentially chosen over DLs as ACL rupture prevention exercises. Here,
in order to shift the hamstring to quadriceps ratio towards the hamstrings, GMs should be favoured ahead of DLs using 50% BW before DLs using 25% BW.

8.2 Background

Strength training exercises such as Deadlifts (DLs) or Goodmornings (GMs) are commonly used in prevention programs for reducing the risk of ACL injury or for rehabilitating low back pain patients, as well as during training to increase an athlete’s specific performance, where the loading conditions play an important role on both the passive and active musculoskeletal structures. Here, exercise kinematics play a key role for governing the lifting mechanics, and therefore modulating the risk of injury% and level of performance (61). In 1999 in Switzerland, the most frequent injuries during fitness training were the shoulder 24.4%, back 16.6%, thigh 11.0% and knee 8.8% (162). The reasons for injury were predominantly attributed to overloading (45.6%) and wrong execution of the exercise (21.1%) (162). Despite these statistics, a complete biomechanical understanding of the loading conditions of many exercises during strength training remains lacking.

The DL is a multi-joint resistance exercise that is performed in a variety of training settings (212). It begins with the lifter in a squat position, with arms straight and pointing downwards, with an alternating hand grip on the bar (61). The movement includes mainly an extension of the knee and hip until the body reaches an upright standing position. The lifting exercise then uses the following muscles: gluteus maximus, erector spinae, hamstrings, quadriceps, trapezius, rhomboideus, deltoideus and finger flexors (79). Due to the fact that the DL is a closed chain exercise (175), it is often used in the prevention of and rehabilitation after anterior cruciate ligament (ACL) reconstruction to improve strength of the muscular structures that surround the knee and hence dynamic stability of the joint (60; 143; 248). The DL is
also one of the three disciplines in powerlifting. The biomechanics of the lift have been studied extensively during competition, focusing on the sumo and conventional styles (61; 60; 151), where the maximal isometric forces in four different positions during DLs were shown to result in a higher potential to increase the force toward the end of the DL (from 3380 to 5829 N) (13). Training using these exercises has also been clearly related to functional adaptation of the spine, where the annual lifted loads of power-lifters has been shown to correlate with the bone mineral content in L3 (80). However, the increased forward trunk tilt during DL lift-off may predispose the spine and back musculature to an increased risk of injury (80; 37). In response to this, Cholewicki and co-workers (37) demonstrated that a more upright trunk at lift-off is able to reduce anterior shear force at the lumbar L4/L5 joint. Furthermore, Escamilla and co-workers (61) showed the importance of keeping the barbell mass as close as possible to the body in order to minimise injury risk to the back as well as to enhance performance. No statistically significant differences were found in this study between the kinematics of high and low-skilled lifters, but they did show differences regarding how the barbell passed the knee: Highly skilled lifters kept the barbell mass closer to the body than less-skilled lifters.

The GM exercise is an assistance movement utilized primarily by weight lifters to strengthen the extensors of the torso, the gluteus, hamstrings and erector spinae (33). Starting in an upright standing position and with the barbell on the shoulders, the hips are progressively flexed until maximum hip flexion is reached, but the knees remained straight throughout. GMs are a good exercise for specifically conditioning lumbar-thoracic flexion and extension of the back (118), but for all level of performance, good lifting technique is required when approaching near maximal effort to avoid acute injury or long-term damage. Here, the low back must have sufficient strength to keep the body in the correct position, since high erector spinae forces are
known to occur, resulting in high shear and compressive forces at the level of L5/S1 (26). The authors stated the importance of sufficiently conditioned lower back musculature and proper sport technique for reducing the risk of back injury (26; 250). In the review of Carpenter DM and Nelson BW (29), the recommendation for low back pain patients was to train using lumbar extension reconditioning exercises with the pelvis stabilized in a specific, progressive and intensive manner, since this was shown to lead to the most favourable improvements in low back strength, muscle cross-sectional area and vertebral bone mineral density. The latter recommendation is in agreement with the general finding that strength training is able to relieve low back pain (119).

During lifting, it is well known that the main part of the axial loading of the spine is due to the large muscle and ligament forces applied over small internal lever arms. Importantly, it is thought that the amount of lumbar flexion (reduction in lordosis) determines the amount of ligamentous involvement in internal loading generation (149; 150), which may or may not be present during heavy lifts (36). Due to the smaller lever arms of the ligaments compared to the muscles, preserving sufficient lordosis when lifting can reduce the bone-on-bone loading between the vertebral bodies due to lower posterior ligament tension (36). However, the preservation of 1-3° margin from full lumbar flexion seems to be sufficient to avoid overloading, and this is consistent with the kinematics observed in highly skilled lifters (36).

GMs and DLs are comparable in their ability to train agility, speed and power in all sport types (182), including typical strength exercises for ACL rehabilitation (139), but also for potential injury risk during exercising (38). Despite the widespread use of GMs and DLs, the critical differences in lower limb and trunk motion, and more importantly the resulting loading conditions on the joints, during GMs and DLs remain unknown. This study
therefore aimed to compare the segment kinematics and joint moments of the lower limbs and spine during the entire lifting action and at the point of deepest flexion during GMs and DLs in the sagittal plane.

8.3 Methods

Nine male and four female subjects with experience in weight training (average age 24.5 ± 4.3 years, mass 74 ± 11 kg, height 180 ± 7 cm) were analysed while performing DLs and GMs exercises. The study was approved by the Ethics committee of ETH Zurich, Switzerland (EK 2012-N-57). One subject provided written informed consent to the publication of their images and all subjects provided written informed consent to participate in the study.

To analyse the motion of the body, an opto-electronic system (Vicon, Oxford Metrics Group, UK) with twelve cameras (MX40) and a sampling frequency of 100 Hz was used. The ground reaction forces were measured using two 400 × 600 mm force plates (type 9281B Kistler, Winterthur Switzerland), one under each foot, with a frequency of 2 kHz. The IfB Marker Set (136), consisting of 55 markers on the legs, pelvis, shoulder and arms, 22 on the back and 2 attached to the barbell, was used (Figure 1). The markers near the spine and on the rear and forefoot had a diameter of 9 mm while 14 mm markers were used on all other segments. Each marker was attached after palpation using double sided skin friendly tape by trained personnel.

Subjects wore their normal shoes for fitness training and shorts, while females additionally wore a bikini top. After a gentle warm-up session running on a treadmill or lifting an unloaded bar, the subjects’ performed the basic motion tasks (136). The subject received standardized instructions for the two exercises (See section Standardised instructions for DLs and GMs). Both exercises started in the upright position. For the GMs, the subjects were advised not to bend their knees. A set of GMs with eight repetitions and an extra load of 25% body weight (BW) on the bar was then performed
Figure 1: Measurement set up including the following: a) subject with the marker set, b) barbell, c) force plates under each foot and d) 1 of 12 vicon cameras.
with the bar positioned on the upper trapezius muscle. Afterwards, the subjects additionally performed sets of eight repetitions each for DLs with 25% and 50% BW on the bar, with the extra loads representing a typical loading of a healthy non-powerlifters and was normalised to percentage of BW.

Standardised instructions for DLs and GMs General instructions:

1. Stand upright with your feet approximately shoulder width apart.

2. Point the feet slightly outward, following the natural divergence of the feet.

3. Lift the thorax to a natural spine position.

4. Hold tension in the core muscles during execution of the exercises.

5. Breathe out during the ascent.

6. Perform the exercise at the same normal speed during the downward and upward movements.

7. Lowest point before turn: No flexion in the lumbar spine.

DL specific instructions:

1. Hold the barbell with a comfortable grip, one hand in a supinated and the other in a pronated position.

2. Keep the head in a horizontal view.

GM specific instructions:

1. Put the barbell on the rear musculus deltoideus and hold it in a comfortable hand position.

2. Keep the head in extension of the spine.
The motion data were reconstructed in Vicon Nexus (version 1.7.1, Oxford Metrics Group, UK). The definition of a repetition of both of the two exercises was based on the start and stop point by using the vertical velocity of the two markers attached on the barbell (vbarbell ≤ 0.04 m/s). The repetitions were time normalized and averaged. In addition, the maximum value for each repetition was averaged. Additionally, the force plates were specifically calibrated to allow for correction of the centre of pressure (COP) (47) and hence maintain accuracy during the inverse approach. The joint centres of the knee and hip were functionally determined from the basic motion tasks (136), and the joint centre of L4/L5 was defined anatomically based on anthropometric data (171). The external joint moments in the sagittal plane were calculated using an inverse approach with a quasi-static solution (249), taking the ground reaction force and kinematic data into account (98), and normalized to BW (140). The flexion / extension moments at the knees and hips were averaged over both limbs. The inverse approach included the position of the joints, the forces acting on each foot, and the gravitational force of the segments (140). Due to slow accelerations of the segments during these exercises, the inertia forces were neglected. All calculations were performed in Matlab (version 8, The MathWorks Inc., Natick, MA, USA).

The position and orientation of each segment was determined relative to the reference segments defined by the standing trial as the neutral position (0° rotation) using a least-squares fit of the corresponding marker point clouds (136). Joint rotations were described using a helical axis approach and for clinically interpretable rotational components, the attitude vectors were decomposed along the axes of a segment fixed, orthogonal, anatomically defined joint coordinate system (136). The joint angles, the curvature and the moments were all analysed in the sagittal plane.

The influences of the extra barbell load (25% BW and 50% BW) and
the type of lift (DLs and GMs) on the maximal segmental angles of the knee and hip, on the maximal and minimal segmental angle of the pelvis relative to lumbar spine and lumbar spine relative thoracic region, on the corresponding ranges of motion (RoMs) of the knee, hip, lumbar and thoracic spines (segmental and curvature approach) as well as on the normalized maximal moments of the knee, hip and L4/L5 were analysed using a multiple repeated-measures ANOVA (with significance defined at $p < 0.05$). Bonferroni adjustment, for the three groups, as well as for the number of parameters, was then conducted to establish significant differences, resulting in significance defined at $p < 0.0033$. All eight repetitions of every subject were averaged for the statistical analyses. Statistical calculations were performed using IBM SPSS software (version 21, SPSS AG, Zurich, Switzerland).

8.4 Results and Discussion

Knee and hip Kinematics:

The observed maximal knee and hip rotations in the sagittal plane, as well as their RoMs, were significantly smaller during GMs than during DLs (Table 1). The typical observed intrasubject standard deviation over eight repetitions of the maximal joint angles of the knee and hip as well as of their RoM was $\pm 2.5^\circ$. These results in the knee were expected due to the type of lifting execution during GMs, where the knee remains almost straight. The smaller flexion movement of the hip during GMs could be a restriction due to the limited length of the two-joint hamstring muscles in the extended knee position. The obtained maximal knee angles during DL were slightly larger than those observed by Brown EW and Abani K (22), while the hip angles remained comparable. No changes in the maximal knee and hip angles or their corresponding RoMs were found between the loading conditions with 25% and 50% BW during DLs. The observed RoMs during DLs of the knee and hip were in agreement with those observed in the study by McGuigan
Table 1: Maximal segmental flexion angle of this an other Studies ((22; 60; 61; 151)) range of motion (ROM) of each segment in the sagittal plane as well as the ROM of the curvature (1/m) of the lumbar and thoracicspine.

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<tbody>
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<td>2.6</td>
<td>2.5</td>
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<tr>
<td></td>
<td>SD</td>
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<td>0.5</td>
<td>0.5</td>
<td>0.4</td>
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<tr>
<td>Lumbal-thoracic</td>
<td>Mean</td>
<td>16.8</td>
<td>18.9</td>
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<tr>
<td></td>
<td>SD</td>
<td>3.8</td>
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<tr>
<td>Hip</td>
<td>Mean</td>
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<tr>
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<tr>
<td>Lumbar-thoracic</td>
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</tr>
<tr>
<td></td>
<td>SD</td>
<td>5.0</td>
<td>5.0</td>
<td>5.0</td>
<td>5.0</td>
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MR and Wilson BD (151).

Kinetics:

During DLs, no changes between the two loading conditions (25% / 50% BW) on the barbell were found in the maximum moment about the knee in the sagittal plane. This seems rather surprising; one might expect higher loading due to the additional load on the bar (Table 2). However, it seems that a slight change of the trunk position is able to considerably modify the sagittal moments about the knee and therefore negate the effect of the extra load. In doing so, the subjects have managed to avoid additional loading at the knees. On the other hand, the hip flexion moment, as expected, increased significantly with additional weight on the barbell during DLs (Table 2).
During GMs, an external extension moment acted at the knee, while during DLs, from a knee flexion angle of $25^\circ$ and higher, the moment produced was a flexion moment (Figure 2). The maximum external knee moments during DLs (Table 2) were comparable to knee moments in the studies of Escamilla et al. (61, 60), although they used much higher barbell loads. Contrary to this finding, the studies of Cholewicki et al. (37) and Brown and Abani (22) showed slightly lower knee moments in comparison to the present study (Table 2).

With the same extra load (25% BW), the sagittal moments about the hip were significantly larger during GMs compared to DLs (Table 2). However, the sagittal hip moments calculated for the current study were 2 to 6 times smaller than the aforementioned studies (61; 60; 37; 22) (Table 2), but this is in line with the reduced barbell loading and was therefore entirely expected.

The largest extension moment at the knee was, in fact, observed during GMs (Table 2). It should be noted that GMs are a rather isometric exercise for the knee flexors, but the hamstrings undergo eccentric and concentric contraction due to motion at the hip (Figure 2). At the same extra weight, the RoM of the hip throughout the exercise was significantly smaller but
Figure 2: Normalized knee moments in the sagittal plane [N·m/BW] (positive for external knee flexion moment) averaged over all repetitions for all subjects during Goodmornings and Deadlifts with corresponding knee flexion angle (°), zero represents a straight leg, defined on the basis of the standing trial) compared to the squat exercise (data taken from (140)). *: Starting point of the eccentric phase at the upright position; blue: DL with 25% extra load; red: DL with 50% extra load; green: GM with 25% extra load; blue dotted: squats with 25% extra load; red dotted: squats with 50% extra load.
the hip sagittal moment significantly larger during GMs compared to DLs (Figure 3). The largest RoMs and the highest sagittal moment in the hip were observed during DLs with 50% extra load (Figure 3).

The motion and loading patterns of DLs were observed to be similar to squats (140). Here, the maximum knee flexion angle was within a few percent (\(\pm 4\%\)) whereas the maximal flexion moment in the knee was higher for squats with 50% extra load (Figure 2). Larger differences were observed in the hip, however, where the maximal flexion moment during DLs was at least 50% larger (Figure 3).

**Relevant outcome for ACL injury prevention:**

A number of studies have identified the force ratio of the quadriceps to hamstring (H:Q) as a risk factor for ACL rupture (54), especially in women (99). Holcomb et al. (99) used, amongst other exercises, GMs and straight leg DLs to modify the H:Q ratio by training within 6 weeks. For the training of multi-joint muscles, such as the M. semimembranosus and semitendinosus, the joint angles of the hip and the knee, as well as their corresponding moments, should be taken into account. Based on the finding that maximal external extension moment in the knee (Figure 2) and a flexion moment in the hip were observed during GMs (Figure 3), it follows that GMs might provide an effective strategy for focused strengthening of the hamstrings. This finding is in agreement with the study of Ebben (54), who demonstrated the importance of hamstring training for the potential reduction of ACL injuries and who further recommended GM training to be included as a preventative measure.

One new and notable outcome of this study is that no larger external knee moments were observed by a larger extra load on the barbell during DLs (Table 2). Compared to squats (140), DLs have the advantage that the flexion moment in the knee is smaller (Figure 2) but the flexion moment in
Figure 3: Normalized hip moment in the sagittal plane [N·m/BW] averaged over all repetitions and all subjects (positive for external hip flexion moment) with corresponding hip flexion angle (°, zero represents a straight hip, defined on the basis of the standing trial) compared to the squat exercise (data taken from (140)). *: starting point of the eccentric phase at the upright position; blue: DL with 25% extra load; red: DL with 50% extra load; green: GM with 25% extra load; blue dotted: squats with 25% extra load; red dotted: squats with 50% extra load.
the hip is larger (Figure 3) using the same extra load. Therefore, based on the observed kinetics and kinematics of the strength exercises, the following ranking is suggested in order to shift the H:Q ratio towards H: GM, DL 50%, DL 25% and squats.

**Back kinematics:**

Neither the RoMs of the lumbar and the thoracic curvatures, nor the maximal and minimal flexion and extension angles of the pelvic-lumbar and the lumbar-thoracic segments, were affected by the execution or the extra weight on the barbell (Table 1).

**Kinetics:**

The flexion moment in the L4/L5 region was significantly higher during DLs than GMs due to the additional load on the barbell (Table 2). However, the two exercises produce the same loading conditions at the L4/L5 region using the same load on the barbell (Table 2). Previously presented data of normalized moments in L4/L5 using 259% BW load during conventional DLs [12] were larger compared to the values of the present study (Table 2). However, the study from Burnett et al. (26) showed smaller moments at L4/L5 during Romanian deadlift exercises compared to the present study, even though they used much higher normalized weights on the barbell (133% BW extra load). During GMs, Burnett et al. (26) found higher moments at L4/L5 compared to the present study, which could be explained by the heavier extra loading on the barbell (72% BW extra load).

**Back training:**

Surprisingly, the relationship between the L4/L5 moment and the lumbar curvature in the sagittal plane was different between the concentric and the eccentric phases of lifting; especially during DLs using 50% extra weight
(Figure 4). During the eccentric phase, the lumbar back maintained its higher curvature longer compared to the concentric phase with the same sagittal moment (Figure 4). Due to the fact that the flexion moments at L4/L5 were similar for the two exercises with 25% BW extra load (Table 2) and no differences in the RoM of curvature or the segmental kinematics of the back were observed (Table 1), from a biomechanical point of view, the two execution types are comparable for the trunk. As a result, differences between the exercises of the kinematics and mechanics in the lower limbs should be considered more relevant.

8.5 **Practical application:**

In order to optimize the training effect of the quadriceps, a large RoM (9) and external flexion moment in the knee is demanded. It follows that the DL is the preferable exercise for quadriceps training, although the extra load did not affect the sagittal moment in the knee. During squatting, similar RoMs in the knee, but higher moments in the sagittal plane and a load dependency of the moments have been observed (140).

To train the M. gluteus maximus, GMs produce a higher sagittal moment but a smaller range of motion than DLs. If a large range of motion is required, DLs are therefore considered the better choice. The small RoM in the knee suggests that GMs should be chosen before DLs at the early stage of rehabilitation for subjects with a previous knee injury. Furthermore, GMs are suited to avoid external flexion moments at the knee. The magnitude of the resulting extension moments during GM is similar to the magnitude of the flexion moment during DLs.

8.6 **Conclusions**

DLs and GMs show different motion and loading patterns for the lower extremities, where the knee remains almost straight during GMs, hence
Figure 4: Normalized L4/L5 moment in the sagittal plane [N·m/BW] averaged over all repetitions and all subjects (positive for external L4/L5 flexion moment) with corresponding lumbar curvature [1/m]. *: starting point of the eccentric phase at the upright position; blue: DL with 25% extra load; red: DL with 50% extra load; green: GM with 25% extra load.
producing a large extension moment. The maximal knee and hip angle, as well as the RoMs of the knee and hip, are smaller during GMs than DLs. Kinematically, the DL is not generally affected by the extra weight on the barbell. The flexion moment at the knee during DLs is also not influenced by the additional 25% load, however, the sagittal moment in the hip is higher during DLs using 50% BW extra load. Based on the higher sagittal moments in the hip and the L4/L5 region with higher barbell loads, great care should be taken to ensure core stability of the trunk during lifting due to high loading of the spine, especially when training with higher extra loads. Finally, for prevention of ACL injuries, GM are recommended for training the hamstrings to quadriceps ratio.

Acknowledgement

We like to thank to Dr. Hans Gerber, Marco Hitz and Peter Schwilch for the technical support.
9 Conclusions & Outlook

Strength training is beneficial for fitness and health for different reasons. However, incorrect execution or overloading of the musculoskeletal system during strength exercises is associated with a small risk of injury (chapter 1).

The goal of this work was to define biomechanically based guidelines for coaches and therapists to execute strength training exercises that provide sufficient mechanical stress on the target structure to induce positive musculoskeletal adaptation without overloading other structures or parts of the body. In the present work, boundary conditions and input parameters for inverse dynamics analysis were gathered and applied to two different squatting and trunk exercises. The approaches that were used combine experimental observations and measurements with advanced numerical techniques to reveal the internal loading conditions in individual subjects, as well as the factors that govern these conditions. This work is thus the first step toward an improved understanding of the influence of different strength exercises on the loading and adaption of the human body.

9.1 Summary of the findings

In chapter 2, the influence of step length and frontal tibial angle on knee and hip joint loading during lunges was analysed. The ROM and loading conditions during lunges were calculated for a combination of step lengths and frontal tibial angles. The resulting data set, including the angles of the knee and hip with the corresponding moments, will allow therapists and coaches to choose a suitable exercise execution based on biomechanical principles for individual patients or athletes.

In order to reduce the risk of injury during strength training but also to provide sufficient stress stimuli on the target muscles, the loading of the knee and hip joints together with the motion of the trunk during squatting exer-
cises were analysed in chapter 3. The squatting exercise is one of the most common strength training exercises and was previously identified as exposing the subject to a potential risk of injury for the knee and the trunk (162). In order to derive the internal loading conditions for this exercise, inverse dynamics analysis based on measured motion and force input data was combined with subject specific modelling. As a critical component for assessing the kinetics (in this case ground reaction forces) in a robust and accurate manner, a new method for calibrating the force plates was developed. In addition to accurate ground reaction forces, the measurement of the motion of the trunk was performed using an advanced and validated marker set. The high level of accuracy achieved using these novel techniques therefore ensured that the successive determination of joint forces and moments allowed an inverse dynamics analysis of the highest quality for assessing loading of the both the knee and trunk. Based on the kinetic and kinematic findings in lunges and squats, no evidence was found to support the common guideline that the knee should not move over the toes. This is in agreement with the earlier findings during squatting (72). In all conditions during lunges, the moments were larger in the rear knee compared to the front knee.

While chapter 3 focused on the sagittal movement, the importance of maintaining proper knee alignment in the frontal plane during strength and conditioning training was presented in chapter 4. Here a significant correlation between the jump performance during competition and the knee varus / valgus position during squatting, drop jumps and imitation jumps was found.

A new method was introduced to analyse the motion of the trunk using optical motion capture with a newly-developed skin marker set (chapter 5). For the first time, this set-up has allowed the loading of the spine to be accurately determined during squatting. As the influence on the accuracy due to soft tissue artefact was unknown, the novel skin marker set was validated
using data from open MRI in a static position (chapter 6). More details on the derivation of the sagittal angles of the spine, including the Cobb angle, are presented in chapter 7. The results of the error analysis demonstrated that the marker set is capable of accurately measuring changes in spine movement, i.e., change of the lumbar curvature. However, large errors were observed in the absolute values of the lumbar curvature. In chapter 5, the validated marker set was used to track changes of spine motion during different executions of the squat exercise. The results of the analysis in chapter 5 will allow biomechanically-based guidelines for therapists and coaches on how to best execute the squat exercise such that soft tissue structures are not overloaded but that the desired musculoskeletal adaptation is achieved. In particular, it was possible for the first time to assess the biomechanical differences between two classic strength training exercises, namely deadlifts and good-mornings, leading to sensible ratings towards guidance for reduced injury risk. Importantly, subjects with disabilities or special training requests may benefit from such findings.

For all inverse dynamics analysis in this work, simple scaling of the reference model based on subject-specific kinematic data was performed. This is a limitation of the current approaches. In particular, the calculation of the moments in the knee and the back using inverse dynamics are rather sensitive to subject-specific variations in the joint rotation axis. In contrast, calculations of hip joint moments are more robust due to the anatomy of the hip. In general, the influence of subject-specific variations on joint alignment is smallest in the sagittal plane, and this is why no transversal or frontal angles have been presented in this work. Currently we are investigating the influence of scaling and weighting of the input data on the error of the modelling and how this error can be limited.
9.2 Practical Applications

Due to the higher moments and a larger ROM, split squats are considered more demanding than normal squats. In particular, the loading of the rear leg has to been taken into account but in general, due to lower extra load on the barbell, the loading of the spine has been found to be smaller. The traditional rule during split squats has been to ensure the leading knee remains behind the toes, resulting in a movement compensation mechanism of the trunk. However, the trunk flexion movement was found to be higher during restricted squats resulting higher stress on the lower back. As a result, coaches and therapists should be advised to supervise the motion of the trunk rather than the forward motion of the knee. Additionally, if the leg muscles are the target of the training, unrestricted squatting is recommended for most subjects. Here the loads on the target muscles such as the quadriceps are higher, the load on the hip is similar, and smaller on the spine. When supervising squats it is important to ensure that proper depth of the squat is chosen due to the high moments on the knee, hip and lower back. The presented differences in kinematics and loading during GMs and DLs therefore allow the appropriate exercise to be chosen, but also provide factors that can now be modulated to target the training according to the data presented. As an example, GMs are recommended to change the ratio between the hamstrings to quadriceps for the prevention of ACL injuries, and the extent of knee flexion can be modulated to provide the required loading.

An additional example of the importance of the lower limb kinematics not only in training but also in the field was demonstrated in ski jumpers, where the virus/valgus alignment of the knee during take-off was strongly correlated with the jump performance during competition. Therefore for these athletes it is important to maintain proper knee alignment in the frontal plane during strength and conditioning training to promote improved kine-
matics during competition. For coaches and therapists, it is crucial to have evidence based knowledge in order to select suitable types of exercises and govern the loads to the target structures without overloading or damaging them or the surrounding soft tissues.

9.3 Transfer of technology

The developed spine marker set has the potential to become a standard tool for assessing the motion of the trunk during optical motion capture in sports research as well as in clinical settings. This work indicated that a change of lordotic/kyphotic curvature is detectable using such non-invasive approaches, but that the absolute amount of curvature cannot be estimated using skin markers alone. Interestingly, the use of the presented back marker set for analysing spinal kinematics seems to be similarly accurate as that of the lower extremities. As a result, the back marker set has seen translation not only for assessing the trunk motion of German gymnasts but also for evaluating the kinematics of ski jumpers during squats, drop jumps and imitation jumps (177). In addition, the marker set is already in use in the UKBB Basel to analyse movement deficits in children with scoliosis (198) or varying leg lengths.

9.4 Impact on teaching

Due to their slow and repetitive motion, strength training exercises provide nice examples for presentation to students. Most of them can be treated as quasi-static, and therefore lend themselves nicely for analysis using inverse dynamics techniques. Furthermore, most students have a good feeling for joint angles, the external loading and the tension in the involved muscles during such exercises. In the bachelor class ”Biomechanics II”, examples of biceps curls and squats are used to discuss the joint reaction forces and the under determined system of muscles. Here, a clear entry point towards mus-
cle optimisation can be given. In the class ”Movement- and Sports Biomechanics”, examples of back training are used to identify the main loading of the spine. Furthermore, two lectures are dedicated towards strength training exercises including inverse dynamics and joint loading. In the master class ”Clinical and Movement Biomechanics”, an entire sequence is dedicated to whole body modelling. Also, a double lecture in the class ”Sports Biomechanics” is dedicated to the different types of loading and strength training exercises.

In order to ensure that the students are sufficiently exposed to the practical elements of kinematic and kinetic measurements, the measurement tools of our motion analysis lab are presented and the students are taken into our lab for practical exercises. To complement this, the master class ”Applied movement analysis” focuses on the possibilities and limitations of pure observation of movement, including strength exercises. Here, a comparison against movement analysis using equipment and measurement devices performed by the students is included. Finally, as guest lectures, the findings of this work are presented practically during the class ”Exercise sciences”.

9.5 Current projects

Design of exercises

Since hip strength weakness is assumed to be associated with several injuries of the lower extremities, one of the key aims in designing exercises is to sufficiently understand the simulated hip strength tasks on the cable machine in order to determine the range of motion and the activity of the M. gluteus medius. The traditional strength exercise using a cable machine to train the hip ab/adductors was simulated using the open source software Opensim (204). Artificial whole body motion data was generated in Matlab. In addition to the neutral movement but also an internal and external rotation in the hip was simulated. An external cable force was applied from different
angles to the foot representing different position of the body towards the
cable machine. The activity as well as the range of motion of the different
parts of the M. gluteus medius was evaluated. Both the angle of the applied
force and the hip rotation position influenced the loading of the M. gluteus
medius. The results suggest that to achieve a higher activity level, different
exercises are required for the anterior and the posterior part. The findings
were presented at the SGS meeting 2016 (Plüss et al.).

Validation of the modelling

The internal loading conditions, especially muscle but also joint contact
forces, in strength exercises could provide strong evidence for adaptation
processes and for improving training and rehabilitation design. The muscle
loading patterns and their general dependency on the joint angles still remain
unknown for many exercises, due to the fact that a direct, non-invasive
measurement is not possible. Musculoskeletal modelling including muscle
optimisation will have a significant impact for answering this type of research
question. Due to a large influence of the subject specific anatomy and tissue
properties, the musculoskeletal modelling itself is challenging but even more
difficult is the validation of this type of computational simulation.

To directly address this problem, we have recently undertaken a series of
measurements that will prove critical for improvement and validations of our
models. As a key part of the on-going so-called "Comprehensive Assessment
of the Musculoskeletal System" (CAMS-Knee project) we aim to evaluate
the accuracy of kinetics and kinematics of the knee during squatting by
comparing musculoskeletal modelling with real in-vivo measurements of the
internal bone-on-bone joint contact forces in six subjects who possess an
instrumented total knee arthroplasty. In this project, six subjects (5m, 1f,
aged 68 ± 5 years, mass 88 ± 12 kg, height 173 ± 4 cm), each with an
INNEX knee implant (Zimmer, CH) were analysed while performing five
repetitions of a squat exercise, performed without additional weight. The tibial component of the implant was instrumented with a 6-axis load cell that could transmit the forces telemetrically. Furthermore, based on the method of the squat study (chapter 3) the motion of the body and the ground reaction forces were measured in a time synchronised manner. Reference Opensim models (204) were individualized according the subject and used to calculate the movement and loading conditions, especially the knee joint contact force, based on the kinetic and kinematic input data. This computed knee joint contact force was then compared with the measured force in the knee implant.

The results now seem to indicate that an overestimation of the knee joint contact force occurs in the Opensim model by a factor of up to 2. The largest differences between the measured and the model knee joint force were observed at large flexion angles of the knee. The large knee angles might also be the reason why the observed differences are higher compared to earlier results in gait, running and stair descent by Heller and Duda (92). No differences could be observed between the concentric and eccentric phases of the movement.

To our knowledge, this is the first time that an Opensim model has been evaluated by means of an instrumented knee implant during strength exercises. Due to higher loading and larger ranges of motion, musculoskeletal modelling seems to be more challenged than when simulating gait or similar activities of daily living. This study suggests that generic models need to be adapted to the subject but also improved to better model the conditions when joints undergo large flexion angles. The initial findings of this ongoing study have recently been submitted to the 34th Conference of the International Society on Biomechanics in Sports (Schellenberg et al.).
9.6 Perspectives and vision

The kinetic and kinematic data gained through inverse dynamics analysis can provide guidelines for coaches and therapists to safely perform strength exercises based on biomechanical principles. It is important that the guidelines for exercises are not only based on experience but is truly based on reliable and accurate scientific evidence. Here, future studies can identify incorrect guidelines such as the "no knees in front of the toes" during squatting and provide reasons for new approaches. It is also becoming clear that the combination of experimentally gathered kinetic and kinematic data together with subject-specific modelling will be a powerful tool in the future. Such an approach would allow not only the estimation of the external loading conditions but also of individual muscle forces. With sufficient validation and widespread application, muscle force calculations during strength exercises in vivo are expected to provide biomechanically based evidence for clinicians and therapists to evaluate and improve training guidelines (196).

In a similar manner, we are not far from using the presented measurement techniques combined with whole body modelling to produce patient-specific models. These models have the power to design exercises that maximise the range of motion and the loading in the target muscles while reducing the overall loading of the body especially on structures that need to be preserved or protected.

Furthermore, the internal forces, moments and joint motion from inverse dynamics analysis can serve as an input, boundary condition or validation of computational models at the tissue level, including mechano-biological models to analyse the biological adaption of human tissue to mechanical loading. In future work, the results of the present work will be coupled with microscopic models, typically built using the finite element (FE) method, to capture changes in the material properties and tissue adaption due to loading.
However, each human body is varies with respect to geometry as well as material properties. It is envisioned that subject-specific models will be developed and used to plan a rehabilitation or training program in the future, allowing safe supervision of the training execution and progress. Future subject-specific, multi-scale models should not only include the macroscopic kinetic and kinematic data presented in this work, but also subject-specific material properties on a microscopic level (30). With sufficient validation and widespread application, muscle force calculations during strength exercises in vivo are expected to provide biomechanically based evidence for clinicians and therapists to evaluate and improve training guidelines (196). In addition to the strong potential for supporting such rehabilitation scenarios, the application of similar models to sports biomechanical research will allow a definition for evidence-based guidelines in strength training for a broad population as well as for individual athletes.
References


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Joint angles of the ankle, knee and hip and loading conditions during split squats


Comparison of the angles and corresponding moments in the knee and hip during restricted & unrestricted squats


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Kinematics and kinetics of squats, drop jumps and imitation jumps of ski jumpers

Carole A. Pauli1, Melanie Keller1,2, Fabian Ammann3, Klaus Hübner4, Julia Lindorfer1, William R. Taylor1 and Silvio Lorenzetti1

1 Institute for Biomechanics, ETH Zurich, Switzerland 2 Department for Sport, Movement and Health, University Basel, Basel, Switzerland 3 Swiss
Ski, Haus des Skisportes, Bern, Switzerland 4 Swiss Federal Institute of Sports, Magglingen, Switzerland

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Kinematics of the trunk and the lower extremities during restricted & unrestricted squats

Renate List, Turgut Gülay, Mirjam Stoop and Silvio Lorenzetti
Institute for Biomechanics, ETH Zurich, Switzerland Journal of Strength & Conditioning Research (2013) 27(6), 1529-1538.

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The spinal curvature of three different sitting positions analysed in an open MRI scanner

Daniel Baumgartner1,2, Roland Zemp1, Renate List1, Mirjam Stoop1, Jaroslav Naxera3, Jean Pierre Elsig4 and Silvio Lorenzetti1 1 Institute for Biomechanics, ETH Zurich, Switzerland
2 School of Engineering, Winterthur, Switzerland
3 Röntgeninstitut Zürich-Al tstetten, Zurich, Switzerland
4 Spine Surgery, 8700 Küsnacht, Switzerland


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**Soft tissue artefacts of the human Back: Comparison of the
sagittal curvature of the spine measured using skin markers
and an open upright MRI**

Roland Zemp\(^1\), Renate List\(^1\), Tugut Gülay\(^1\), Jean Pierre Elsig\(^2\),
Jaroslav Naxera\(^3\), William R. Taylor\(^1\) and Silvio Lorenzetti\(^1\)
\(^1\) Institute for Biomechanics, ETH Zurich, Switzerland
\(^2\) Spine Surgery, 8700 Küsnacht, Switzerland
\(^3\) Röntgeninstitut Zürich-Altstetten, Zurich, Switzerland Plos One (2014)
9(4), e95426.

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**Kinetic and kinematic differences between deadlifts and good-
mornings**

Florian Schellenberg \(^1\), Julia Lindorfer \(^1,2\), Renate List\(^1\), William
R. Taylor\(^1\) and Silvio Lorenzetti\(^1\) \(^1\) Institute for Biomechanics, ETH
Zurich, Switzerland

\(^1\) Technikum Wien, Höchstädtplatz 6, 1200 Wien, Austria

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CURRICULUM VITAE

Date of Birth: September 28, 1974
Place of Birth: Schaffhausen, Switzerland
Married, daughter: Samira, February 14, 2014

Institutional Appointments
Since 2014 Permanent Senior Scientist at the Institute for Biomechanics,
ETH Zurich
2007 - 2013 Lecturer and Senior Research Associate at the Institute for
Biomechanics, ETH Zurich
2004 - 2006 Ph.D. Student and Assistant at the Institute for Biomechanics,
ETH Zurich
2003 Post-doc, Department for Space Research and Planetology,
University of Bern
2002 - 2013 Educator for the Star Education, School for Training and
Recreation
2000 - 2003 Ph.D. Student at the Institute for Physics, University of Bern
1999 - 2003 Assistant at the Institute for Physics, University of Bern

Education
2007 Dr. sc. ETH Zurich
New method to determine the Young’s modulus of single trabeculae
2003 Dr. phil.-nat. UNIBE (Department Space Research and Planetology)
Auswurfalter und Bestrahlungsgeschichte der Meteorite vom Mond, Mars und Asteroiden anhand von Edelgasisotopenanalysen

2000 Master in Physics (Mathematics, Astronomy), University of Bern, Switzerland
1998 Exchange term, University of Strathclyde, Glasgow, UK
1995 - 2000 Studies in Physics, University of Bern, Switzerland
1994 Matura Type C (Natural Sciences), Kantonsschule Schaffhausen, Switzerland

Further Education
2014 Advanced course "Good Clinical Practice (GCP)", Modul 3, CTC, Universitätsspital Zürich
2014 Basic course "Good Clinical Practice (GCP)", Modul 1+2, CTC, Universitätsspital Zürich
2013 Course "Project Management Leadership Skills", ETH Zurich
2010 Certificate "Teaching at ETH Zurich", teaching at master’s degree level
2009 Certificate "Teaching at ETH Zurich", teaching undergraduate courses
2008 Venture Challenge
2005 Special module fitness, Physical Education, ETH Zurich

Awards and Honors
2014 Finalist Spark Award, top 20 most promising invention, ETH
2013 Swiss Olympic Sport Science Award, third prize
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**Editorial Boards**

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Other activities

*Competitive Sports & Strength and Conditioning Expertise*

2009 - 2012    Swiss Sailing Championship Dolphin 81 with SUI 55
2004          Vice Swiss Champion, up to 90 kg, Swiss Drug-Free Powerlifting Federation
2002          Swiss Champion, up to 82.5 kg, Powerlifting, SPC
2009 - 2011    Head of the expert commission for "Fachwissen", SFCV, exam
               fitness instructor with Federal Certificate, Bundesamt für Bildung und Technologie (BBT)
2007 - 2011    Expert at the federal exam for "Fitnessinstructor with Federal Certificate" (BBT)

*Expeditions*

06/07          ANSMET, NASA expedition to Antarctica to recover meteorites
02/02/05/09    Expedition in the Omani desert to recover meteorites

*Hunting*

Since 2014     President JagdSchaffhausen
Since 2011     Expert Jägerprüfungskommission Kanton Schaffhausen
2008-2013      Vice President and Webmaster JagdSchaffhausen
2006 - 2011    Treasurer JagdSchaffhausen
2007 - 2015    Trained hunting dog Gwendi
Since 2015     Hunting dog Centi
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