Comparing mechanical discomfort and risk of low back pain or injury when wearing load carriage systems

Author(s):
Wettenschwiler, Patrick D.

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COMPARING MECHANICAL DISCOMFORT AND RISK OF LOW BACK PAIN OR INJURY WHEN WEARING LOAD CARRIAGE SYSTEMS

A thesis submitted to attain the degree of
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(Dr. sc. ETH Zurich)

presented by
PATRICK DANIEL WETTENSCHWILER
MSc, ETH Zurich
born on 05.05.1984
citizen of Rapperswil-Jona SG, Switzerland

accepted on the recommendation of
Prof. Dr. Stephen J. Ferguson, examiner
Dr. Silvio Lorenzetti, co-examiner
Dr. Simon Annaheim, co-examiner

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Summary

Load carriage has become an everyday task for a large part of the population. Special attention is required in the fields of school children’s load carriage, infantry, or recreational outdoor activities, as the typical loads to be carried in these fields have increased over the last decades. The increasing loads call for a constant improvement of the type and design of load carriage systems. The biggest challenges due to the mechanical interaction between a load carriage system and the human body concern the discomfort and the risk of low back pain or injury.

Discomfort is a subjective perception depending on biomechanical factors like soreness, numbness, stiffness, or other comparable perceptions and it can directly influence the user acceptance of a load carriage system. Today, discomfort is usually assessed with subject studies, but they are labor-intensive, costly, and often include bias. Therefore, the first goal was to provide a first validated physical model to assess the mechanical aspects of discomfort when wearing load carriage systems. The validation required the collection of new in vivo data, which in turn called for an improved measurement technique regarding the pressure distribution at the interface between load carriage system and body.

The second goal of this thesis focused on the risk of low back pain or injury. Low back pain or injury is a serious health issue that concerns more than 50% of the adults. Regarding load carriage, known risk factors like the peak shear and the cumulative compression forces in the L4-L5 intervertebral disc must be minimized in order to decrease the risk of low back pain or injury. As these parameters are ethically impossible to measure in vivo, models are necessary to evaluate the risk of low back pain or injury. While existing numerical models already offer the prediction of the loads on the lumbar discs for different postures or everyday activities, they usually do not model the complex interaction with a load carriage system and the relevance of subject-specific lumbar curvature that is adapted to the load condition is unknown. Therefore, the second goal was to present an approach to efficiently model the effect of load carriage systems on the body in numerical simulations and to investigate the relevance of precise lumbar curvature data in spinal models.
In a first subject study, the reliability of Tekscan pressure sensors was assessed and possibilities to improve their reliability were investigated. These results allowed the identification of mechanical predictors of discomfort during load carriage in a second subject study. Through comparison of these mechanical parameters from the second subject study with the corresponding parameters measured on an instrumented dummy, a validation of that dummy was conducted. For the second goal, the same instrumented dummy was used in an approach to compare the risk of low back pain or injury during load carriage by combining physical and numerical modeling. The dummy provided the external forces resulting from the mechanical interaction between the body and a load carriage system, while the numerical model calculated the compression and shear forces in the lumbar discs. An existing validated rigid body model was used for the numerical modeling. A third subject study was conducted to provide the lumbar posture input data and determine its relevance.

Regarding the first goal, the instrumented dummy could be validated to accurately simulate the interaction between the human body and load carriage systems. It can explain 75% or more of the variation in discomfort. This dummy represents the first validated model to assess discomfort during load carriage and it exhibits a higher predictive power than previous models. Regarding the second goal, a novel approach of assessing the risk of low back pain or injury during load carriage was presented by combining physical and numerical modeling. Additionally, the necessity of precise lumbar curvature data as input for spinal models was shown for the realistic prediction of shear forces.

The presented models provide much needed opportunities for future research and development of load carriage system type and design. They enable hundreds of measurements in a reliable and controllable laboratory environment and at reduced cost, which provides freedom to investigate new or unconventional concepts. Both manufacturers and users of load carriage systems will profit from future improvements regarding discomfort and risk of low back pain or injury.
Zusammenfassung


Das zweite übergeordnete Ziel war es daher, eine Methode zu erstellen, um das Verletzungsrisiko im Lendenwirbelsäulenbereich bei Tragsystemen zu beurteilen und die Relevanz von präziser Lendenwirbelsäulenkrümmung in Modellen zu untersuchen.


1. Introduction

1.1. Motivation

Different cultures have developed a large variety of load carriage systems like yokes, satchels, head baskets, backpacks, double packs, vests and many more. Over time, load carriage systems have been adapted for specific fields of application like school children’s load carriage, recreational activities, or infantry. During the last decades, loads carried in load carriage systems have increased markedly [1, 2]. These increasing loads constantly demand for an improvement of the load carriage systems with respect to two different aspects: the subjective discomfort and the risk of injury.

The subjective discomfort is an important factor with considerable influence on the user acceptance of a load carriage system [3, 4]. It depends on biomechanical factors that are responsible for feelings of soreness, numbness, stiffness and other comparable perceptions [5, 6]. Comfort, in contrast, is considered a ‘continuous dimension of experience – varying from strongly positive (very comfortable) to strongly negative (very uncomfortable)’ [7]. Another definition explains comfort simply as the absence of discomfort and, for testing purposes, proposes the measurement of discomfort rather than the measurement of comfort [8]. Within discomfort, thermal aspects may also play a role, namely temperature and humidity, but these aspects are well understood [9, 10]. This thesis therefore focuses on the mechanical aspects of discomfort.

Most studies investigating mechanical discomfort during load carriage rely on subject trials, in which the discomfort is directly assessed in questionnaires [11-14]. Since subject studies are time consuming, expensive, and highly subjective, it would be desirable to have a physical model that allowed the comparison of load carriage systems regarding mechanical discomfort with quantifiable outcome parameters. Stevenson et al. [15, 16] designed such a model for the mechanical evaluation of load carriage systems, but their model lacks a solid validation. Their model uses the mechanical parameters that characterize the effect of a load carriage system on the human body to estimate discomfort. However, these mechanical parameters have never been measured on subjects. A direct comparison between the model and human subjects to examine how closely the model represents reality was therefore im-
possible. It is also unknown, which of the mechanical parameters, when measured on subjects, would be significantly associated with discomfort during load carriage. An obstacle that may have prevented previous researchers from measuring the mechanical parameters that characterize the effect of a load carriage system on the human body is the lack of appropriate pressure sensors to assess the pressure distribution at the interface. There are existing sensors that allow pressure measurements on flat surfaces, but their reliability in measurements on topographically complex surfaces like the human torso are unknown. The first goal of this thesis was to fill these knowledge gaps in the pressure measurement technique, the in vivo data of parameters that characterize the mechanical interaction between load carriage system and body, and their value in predicting discomfort until a solid validation of an instrumented dummy to assess the mechanical aspects of discomfort during load carriage was possible.

Regarding the risk of injury involved in load carriage, the variety ranges from simple foot blisters or pain in the lower extremities to more serious medical issues like the occurrence of low back pain or injury or the damage of the brachial plexus [17, 18]. The influence of the type and design of a load carriage system on the risk of injuries in the lower extremities is limited. In contrast, the risk of damage of the brachial plexus is more directly related to the load carriage system type and design due to the pressure underneath the shoulder straps [19]. Despite the severity of this type of injury, it is not the biggest concern of load carriage system users, as the occurrence of an actual deficiency of the brachial plexus due to load carriage is rare and, if treated appropriately, also remediable [20, 21]. This is different in the case of low back pain or injury: once this type of injury occurs, a later reoccurrence is much more likely [22, 23]. Of all the possible risk of injuries related to load carriage, this thesis therefore prioritized the risk of low back pain or injury.

Generally, low back pain is already a major health issue in our society: In Switzerland, 40% of the adults suffered from back or low back pain during the four weeks prior to being asked [24]. According to Wieser et al. [25], low back pain is the ‘most prevalent health problem in Switzerland and a leading cause of reduced work performance and disability’. Regarding load carriage, the increasing weights make low back injuries an even more serious issue [26-28]. In order to prevent low back pain in manual lifting tasks, several limits for compressive and shear forces in lumbar discs
have been proposed in literature. Waters et al. [29] proposed a maximal spinal disc compression force of 3400N at the L5-S1 vertebral disc level. Gutierrez et al. [30] restricted a study to female workers and recommended a limit for compressive lumbar spine forces of 2800N, as well as a limit for spinal shear forces of 300N. So far, in current literature, no limit has been proposed for spinal torsion. All these absolute values have little predictive power, but a study by Kerr et al. [31] reported peak lumbar shear force and cumulative lumbar disc compression at the L4-L5 level as biomechanical risk factors for low back pain. These risk factors allow a comparison of different load carriage systems and designs on a more detailed scale than the discrete values previously reported. Due to ethical reasons, forces in the lumbar discs are nearly impossible to measure in subject studies. Instead, numerical simulations are a good alternative. Already existing multibody models or finite element simulations allow the calculation of compression and shear forces in the lumbar spine, e.g. [32-35], but there are two major challenges to the development of accurate models that simulate load carriage: the forces and moments resulting from the mechanical interaction between load carriage system and body have to be known and the lumbar curvature has to be defined. Therefore, the second goal of this thesis was to present a novel approach of calculating the loads in the lumbar spine during load carriage by combining physical and numerical modeling and to investigate the relevance of lumbar curvature data in spinal models.

1.2. Specific Objectives

A first knowledge gap was to be filled by determining and potentially increasing the reliability of sensors that can measure the pressure distribution at the interface of the human body and load carriage systems. The new insights gained in that study then enabled a second subject study, in which the previously unknown mechanical predictors of discomfort during load carriage were identified. This newly gained knowledge then allowed the validation of an instrumented dummy through comparison of the mechanical parameters from the subject study with the corresponding parameters measured on the dummy. Regarding the risk of low back pain or injury during load carriage, a solution was needed to the above mentioned challenges to the development of accurate models that simulate load carriage: A novel approach to compare the risk of low back pain or injury during load carriage was presented by combining physical and numerical modeling. The physical model (the dummy that was already
used to assess discomfort) measured the forces and moments acting on the human body by a load carriage system and delivered these data as input for the numerical model. A third subject study was necessary to provide the lumbar posture input data and determine its relevance for the numerical model.

Summarized, the thesis focuses on discomfort and the risk of low back pain or injury during load carriage and consists of the following four aims, whose interaction is displayed in Fig. 1:

**Aim 1**: “Determining the reliability of Tekscan sensors on the body surface of human subjects wearing load carriage systems.”

**Aim 2**: “Identifying mechanical predictors of discomfort during load carriage.”

**Aim 3**: “Validation of an instrumented dummy to assess the mechanical aspects of discomfort during load carriage.”

**Aim 4**: “A novel approach to assess the risk of low back pain or injury during load carriage using numerical and physical modeling.”

**Fig. 1. The interactions of the four aims of this thesis.** The first three aims are focused on discomfort (blue) and the fourth aim is focused on the risk of low back pain or injury (red).
1.3. Thesis Outline

Chapter 2 contains the scientific background of the thesis. Firstly, the status of the current knowledge on the mechanical aspects of discomfort during load carriage is presented. Secondly, the complexity of spinal loading during load carriage is discussed, including postural and muscular adaptations, as well as their kinetic consequences.

Chapter 3 reveals the intra-subject test-retest reliability of pressure measurements using Tekscan sensors on the body surface of human subjects wearing load carriage systems. It further provides specific data processing steps that increase the reliability and also analyzes the influence of the sensor location, the type load carriage system, and the activity of the subjects. This information is crucial for an accurate measurement of mechanical predictors of discomfort in the following chapter.

Chapter 4 identifies objectively measured mechanical predictors of discomfort during load carriage. The results of this subject study include mechanical predictors that are specifically intended for load carriage systems of the type backpack with hip belt, as well as predictors that can be applied on any type of load carriage system. Additionally, recommendations for users are given in order to minimize the spinal loading without decreasing the overall discomfort.

Chapter 5 presents for the first time the validation of an instrumented dummy to assess mechanical aspects of discomfort during load carriage. The mechanical predictors measured in the subject study of chapter 4 are compared with the parameters that resulted from the corresponding measurements on the dummy. Their correlation provides the validation of the dummy. Its predictive power is assessed through multiple linear regressions with the discomfort values from the subject study as independent variables.

Chapter 6 proposes a novel approach of calculating the loads in the lumbar spine and thus the risk of low back pain or injury during load carriage by combining physical and numerical modeling. The external forces acting on the human body from the interaction with the load carriage system are measured on the instrumented dummy that has been validated in chapter 5. These forces serve as relevant input data for an already existing validated rigid body model. The remaining input data, the in vivo
lumbar posture during load carriage, are gathered through upright MRI measurements with subjects wearing a typical load carriage system in three different load conditions.

**Chapter 7** is a synthesis of the results of the previous chapters in this thesis. It summarizes the individual studies and their main outcomes, addresses their scientific and societal benefits, their limitations, and presents an outlook for future research.

### 1.4. References


2. Background

2.1. Mechanical Discomfort during Load Carriage

Most of the currently available literature about the mechanical aspects of discomfort during load carriage consists of subject studies. They are typically limited on investigating one or two specific aspects of discomfort during load carriage. Past findings include investigations about the effects of different materials, the weight of the load, the region of discomfort, the duration of load carriage, or the center of mass. The investigated load carriage systems and the populations represented by the respective subject samples differ widely, yet the previous findings still offer valuable insights into individual aspects of load carriage.

Martin et al. [1] compared different materials in the shoulder straps of backpacks and found that a monofilament air mesh performed better than a foam, distributing the pressure over a larger area and providing lower peak pressure values, with the additional benefit of an increased heat loss function. Another study from Birell et al. [2] assessed the skeletal discomfort after a one hour march with 24kg total load distributed in a combination of webbing and backpack. They investigated the lower back region as well as the lower extremities and found that the foot was rated to be the most uncomfortable region. Additionally, they found females to perceive significantly more discomfort in the hip region than males. The effect of the duration on discomfort was shown by Simpson et al. [3], who proved that with increasing duration of walking with a backpack, the participants reported significantly increased shoulder, neck and upper back discomfort. Park et al. [4] reported that more weight led to more discomfort in a study with a combined ballistic and load carriage vest. Discomfort was even higher, if the load was distributed asymmetrically.

Regarding the vertical position of the load with respect to the body and its influence on discomfort, there are two interesting studies: Simpson et al. [5] loaded female hikers with a backpack containing 30% of their bodyweight. The center of mass inside the backpack was tested in three different vertical positions. The high load was the most preferable condition, but the result was not significant. Devroy et al. [6] compared thoracic vs. lumbar placement of a backpack, while the center of mass was constant inside the backpack. The subjective scores indicated a preference for the
lumbar placement, but these results were not significant either. Compared to the most common European load carriage systems like backpacks and vests, the most extreme difference in the vertical position is given in a traditional African load carriage technique, the head-loading. However, according to Lloyd et al. [7], head-loading is characterized by significant neck pain. This is also the case after long-term habituation. Compared to a backpack, the level of neck pain during head-loading is so high that it does not compensate for the reduction of discomfort in most other areas of the body.

In order to summarize the findings about discomfort during load carriage, Golritz et al. [8] wrote a review in 2011, focusing on the effect of load carriage system weight, design, and placement, but a quantitative outcome was not possible. Instead, the main outcome reports conflicting evidence e.g. on the effect of load positioning and designs. Most importantly, this review reported that the majority of the included studies have an insufficient methodological quality and cannot be fully relied upon. The majority of the studies were criticized for either lacking a sample size justification, lacking a training session for examiners, or the use of measurement instruments without providing information about their calibration, validation, and reliability. Disadvantages of subject studies further include challenges such as intra-subject and inter-subject variability.

An alternative to subject studies would be the use of physical models to assess the discomfort of load carriage systems. Such a model has been developed by the Ergonomics Research group at Queens University in Canada [9, 10]. Their model measures mechanical parameters that characterize the effect of load carriage systems on the body and associates them with discomfort that has previously been quantified in a subject study [11]. E.g. for the shoulder area, the found a coefficient of determination $R^2 = 0.31$ between discomfort and average pressure at the interface with the load carriage system [11]. Compared to subject studies, such a physical model offers several advantages: Large numbers of comparisons enable the systematic investigation of effects of load carriage system type and design. At the same time, subjective bias resulting from aspects like the personal mood of the subjects, their fashion preferences, or their individual physical condition and fatigue level can be excluded. However, the biggest challenge involved in the use of a physical model for the measurement of a subjective perception like discomfort is the validation of the
model. This is where the model developed at Queens University is limited, as for a solid validation, the correlation between the mechanical parameters measured on the model and the same mechanical parameters measured on subjects need to be known. This correlation was not evaluated, because the mechanical parameters in question have never before been measured on human subjects. A technical obstacle may be the reason for this lack of in vivo data, as the currently available pressure distribution sensors are not built for use on a soft and topographically complex surface like the human body.

2.2. Spinal Loading during Load Carriage

In the past, a common attempt to reduce the risk of low back pain or injury due the manual handling of loads has been to define recommended limits. According to the International Standard on manual handling [12], a limit of 25kg for the mass of an object in non-repetitive manual lifting in professional use protects 95% of the male adult working population and 70% of the female working population from musculoskeletal disorders. Lifting is hereby defined as moving an object upwards without mechanical assistance and could be applied to the task of lifting up a load carriage system to wear it. Apart from the mass of an object being handled, the risk of musculoskeletal disorders depends on several other factors like the size of the object, the working posture and the frequency and duration of manual handling [12]. The discussed International Standard also provides a limit for carrying, an activity that would be more useful in defining limits for load carriage systems. However, the definition of the carrying limit uses a cumulative mass related to a carrying distance and frequency and suits a repetitive task rather than a continuous carrying process as it occurs in the use of backpacks or other load carriage systems. In any case, a general mass limit may be the easiest solution of implementing a recommended limit, e.g. in an industrial environment, but it is a very unreliable tool to restrict the load on the spine. To really understand the mechanical interaction between externally applied loads and the loading of the spine, it is important to know how the human body behaves during load carriage. This includes the occurrence of postural adaptations or changes in the muscular activity as a result of load carriage.
2.2.1. Postural Adaptations

Due to the evolution in gravity, the shape of the spine in a natural upright position without external load in a healthy human must be regarded as the optimal position, in which the structures like vertebral bodies, intervertebral disks, surrounding muscles, ligaments, and tendons as well as the neural system are in the best configuration to bear vertical loading and to maintain stability in case of sudden perturbations. A deviation of the spinal posture in the frontal or sagittal plane may induce problems if maintained for a longer period of time. Nevertheless it can be necessary to adapt the spinal posture in order to save energy or to decrease momentary peak forces and moments. It is also possible that the spinal posture is altered because muscles are not strong enough or do not have the right lever arm to maintain the natural position of the spine under load.

Most load carriage systems are symmetrically designed in the frontal plane, but if a system is carried over one shoulder only, the symmetry is lost. Bettany-Saltikov et al. [13] found a significantly increased thoracic lateral curvature with a backpack of 17% body weight if carried over one shoulder compared to both shoulders. Negrini et al. [14] tested school children wearing 8kg backpacks, also corresponding to 17% of their body weight, and they found that the asymmetrical load produced a modification in all anatomical planes, whereas the symmetrical did not. However, a pilot study conducted by Chansirinukor et al. [15] investigated the cervical posture of school children and reported it not conclusive that carrying a backpack unilaterally alters cervical posture. A reason for this might be the relatively small weight of the backpacks, which were only 5.2kg, corresponding to approximately 10% of the children’s body weight. Fowler et al. [16] showed that unilateral carrying of Royal Mailbags led to lateral deviation in the lumbar region. Another asymmetrical load carriage system was studied by Pascoe et al. [17], who revealed that one-strap athletic bags promoted lateral spinal bending and shoulder elevation, while two-strap backpacks significantly reduced these effects. The majority of these studies thus supports the assumption that asymmetrical loading is promoting an unphysiological postural change of the spine in the frontal plane.

With increasing weight of load carriage systems, asymmetrical loading is usually out of the question and the sagittal curvature of the spine is the main concern. Wearing a backpack shifts the overall center of mass posteriorly if no adaptation in posture.
takes place. Most studies demonstrate a forward lean of the trunk while wearing a backpack [6, 17-25]. Only Al-Khabbaz et al. [26] reported backward lean in university students wearing backpacks loaded with 10%, 15%, and 20% of body weight. The setup of Al-Khabbaz et al. had the subjects standing still, but distinguishing between static and dynamic conditions is not sufficient to explain the discrepancy, because many of the studies mentioned above showing a forward lean of the trunk also contained a static setup. Differences may originate from variation in backpack design or location of the backpack’s center of mass, but the large majority of the studies indicate that a forward leaning of the trunk is the typical reaction to carrying a backpack.

Mrozowski et al. [27] observed a lowering of the overall center of mass (body and load) by around 4cm with a 20kg backpack and they assumed a decrease of the radius of the physiological curves of the spine and of the height of intervertebral discs to be responsible. The change in the craniovertebral angle during backpack use was measured in a number of studies and they all detected a decrease, indicating a more forward and downward position of the head [18, 28-32]. The forward displacement of the head helps to compensate the backward shift of the overall center of mass due to backpack loading. In addition, an increase in cervical lordosis was found while wearing backpacks [19, 33, 34]. This is in accordance with a decrease of the craniovertebral angle under the assumption that there is no rotation of the head in the sagittal plane. Looking at the thoracic kyphosis with the use of backpacks, current literature is not conclusive. Orloff et al. [35] and Chow et al. [19] showed an increase, while Chow et al. [36] and Bettany-Saltikov et al. [13] found a decrease. There is less doubt regarding lumbar lordosis, where a decrease with backpacks is reported in all involved studies [19, 33, 37].

Three studies investigated postural effects of a front pack or a backpack worn anteriorly: Fiolkowski et al. [38] measured a smaller forward head position with the front pack than with the backpack, Chow et al. [39] detected less spinal deformation with a backpack worn anteriorly than with posterior carriage, but according to Wang et al. [33] anterior backpack carriage induced a significant increase of thoracic kyphosis. Differences in the design of the backpacks and the amount of load included may well be responsible for these variations in the effects of anteriorly worn backpacks or front packs. The effects of a backpack and a double pack with a third of the load in front have been compared by Kim et al. [31] and they revealed that when carrying a dou-
ble pack, the forward head angle and forward head distance decreased compared to carrying a backpack. These findings indicate that by at least placing part of the load in front, the postural deviation could be reduced.

2.2.2. Change in Muscle Activity

Muscular activity changes when wearing a load carriage system and this can be measured by electromyography. The activity of the rectus abdominis can increase with the use of a backpack [6, 26, 40]. However, Hong et al. [41] found no increase in the rectus abdominis activity with backpack loads of 10%, 15% and 20% body weight. Erector spinae activity decreases with backpack use [6, 40, 42, 43], but may also increase with heavy loads in the range of 30 - 50kg [44, 45]. Decreasing activity of erector spinae makes sense considering the external moment that is exerted on the body by the backpack in an upright condition. Thus for moderate loads, erector spinae activity is reduced, but the rectus abdominis has to compensate the additional moment. For heavy loads, increasing erector spinae activity may indicate a need to enhance spinal stability, especially if an extended forward lean of the trunk occurs. Erector spinae activity is then necessary to inhibit further spine flexion, while rectus abdominis activity can be reduced. Apart from a muscle activity increase to compensate an existing external moment, a preventive co-contraction can become necessary with increasing load. This may serve the purpose of stabilization against moments on the spine that suddenly occur if a heavy load moves out of balance. Such co-contraction is always accompanied by an increase in spinal compression. Conforming results are given by Harman et al. [45], who presented a similar activity of spinal erectors with backpack loads of 6kg, 20kg, and 33kg, but double activity with 47kg. Similarly, Al-Khabbaz et al. [26] found a disproportionally increased rectus abdominis activity as backpack load increased. A compensation of the external moment has also been observed with front packs. Front load increases activity of the erector spinae and decreases rectus abdominis activity [40]. Kim et al. [31] noticed an increase in activity of upper trapezius, sternocleidomastoid and midcervical paraspinal muscles with a backpack and increased sternocleidomastoid muscle activity with a double pack. Regarding the trapezius muscles, Knapik et al. [44] stated that framed packs with hip belts reduce the muscle activity. These findings indicate the complexity of the muscular activity during load carriage.
2.2.3. Kinematic Consequences

Spinal loading due to load carriage systems depends on several factors like external forces and moments exerted on the body, changes in internal forces and moments resulting from postural adaptations, as well as changes in internal forces and moments resulting from adjusted muscle activity. It must be noticed that for dynamic situations, relative movement between body and load may additionally alter forces and moments acting on the spine. Ren et al. [46] have been dealing with this problem and they showed that an elastic backpack suspension can reduce peak values of ground reaction forces and lower limb joint loads, while creating a larger relative motion between body and load. Because Ren et al. [46] suspected affection of the user’s balance and agility, they suggested looking for the right combination of stiffness and damping in backpack suspension, in order to allow a compliant suspension without excessive relative motion. To this point, no such systems have been investigated.

Due to ethical reasons, it is almost impossible to measure spinal disc compression forces in humans in vivo. Alternatively, an indicator of the compression force is the decrease in spinal disc height during or after loading and this has been measured in various studies: Malko et al. [47] investigated the recovery of disc volume after unloading and detected an average volume increase of 5.4% three hours after removing a 20kg load. Hutton et al. [48] measured an average gain of 11% of the volume (or 0.90cm$^3$ of fluid per disc) in lumbar intervertebral discs during an overnight bed rest. Furthermore, wearing a backpack with 40% body weight for four hours produced a volume decrease equivalent to the decrease in volume usually obtained at the end of the day. Neuschwander et al. [49] reported that disc compressibility is greater in more caudal lumbar intervertebral discs. Goh et al. [25] used a biomechanical model to calculate the forces in the lumbosacral spine from ground reaction forces. With a backpack of 15% and 30% body weight, respectively, they calculated an increase in the lumbosacral force of 27% and 64%, respectively, when compared to walking without backpack. However, the model of Goh et al. [25] is lacking adaptations in the lumbar curvature, as a fixed orientation of the lumbosacral joint is assumed relative to the trunk. Additionally, their model simplifies the trunk muscles into a single back extensor muscle and a single flexor muscle with fixed moment arms.

To understand the influence of postural adaptations and muscle activity on spinal compression, it is helpful to also discuss the center of mass of load carriage systems.
Regarding spinal compression, the optimal situation to carry load is if the center of mass of the load is close to the center of mass of the body in the transverse plane. This minimizes the external moment of the load on the spine. As soon as increased muscle activity is necessary to compensate an external moment, the compressive force on the spine increases as well, unless the antagonist activity can be decreased accordingly at the same time. The further away in the transverse plane the center of mass of the load is from the center of mass of the body, the higher is the external moment. Following this logic, front packs would therefore be worse than backpacks regarding the increase in spinal compression due to adapted muscular activity: The lever arm of the erector spinae is smaller than the lever arm of the rectus abdominis, so to generate the same moment, the erector spinae has to produce a higher force. Chow et al. [50] support this concept by reporting that 20 minutes of anterior carriage of a backpack with a load of 15% of body weight led to a higher loss in spinal length than with posterior placement of the backpack. A preventive co-contraction of the erector spinae and the rectus abdominis to stabilize the spine against suddenly occurring moments from heavy load slipping slightly out of balance, undoubtedly raises the compressive force on the spine. A good opportunity to reduce the load on the upper torso and on the spine is to impose a substantial amount of load bearing onto the hip by using a stiff suspension system in combination with a hip belt [51-54].

There is very little data available in literature about shear forces or torsion and bending moments in the spine. The reason for this may be that today, the results available from in vivo measurements only report the intradiscal pressure or the total force [55-58]. As a consequence, most models that can predict spinal loads typically lack a solid validation for the calculation of shear forces and moments. Nevertheless, the importance of shear forces in spinal loading during load carriage is given: Gutierrez et al. [59] found a Spearman correlation coefficient between the prevalence of low back disorders and peak shear force of \( R = 0.9 \) (\( p<0.005 \)) studying female workers. According to Gutierrez et al. [59], a shear forces limit of 300N would protect 90% of the female workers. Independent of the gender, McGill et al. [60] suggest a recommended upper limit of 500N for shear forces, but as already discussed above, such discrete force limit recommendations are rather critical. The failure tolerance of a spinal disc decreases over time during constant loading, but also during repetitive cycles of loading [61]. This means that an injury can occur under load conditions that have
been largely unproblematic beforehand. Kerr et al. [62] did not recommend a discrete force limit, but instead identified the peak shear force and the cumulative compression force as a biomechanical risk factors for low back pain. Similar results have previously been reported by Kumar et al. [63]. Therefore, in order to successfully reduce the risk of low back pain or injury, the goal must be to reduce these known risk factors as far as possible, regardless of a discrete limit value.

One way to predict the loading of the spine during load carriage, while taking into account the effects of postural adaptations and muscular activity, is to use numerical models. However, there are two major challenges to the development of accurate models: the forces and moments resulting from the mechanical interaction between a load carriage system and the body have to be known and the lumbar curvature has to be defined. Regarding the first challenge, modeling the interaction between load carriage systems and the human body, e.g. in multi-body models, would be very labor-intensive. For each load carriage system of interest, the individual stiffness and damping properties that characterize the interaction between the system and the body at all contact points would have to be assessed in order to model these elements appropriately. Additionally, validation experiments and simulations would be necessary. Regarding the second challenge, capturing postural information like lumbar curvature data is mostly done using skin markers [64, 65], despite the fact that relative motion between the skin and the underlying bones can introduce errors [66]. In light of the inaccuracy related to this common method of measuring the lumbar curvature, it would be very interesting to know, whether the lumbar curvature adaptations during load carriage significantly altered the lumbar compression and shear forces, which are the relevant parameters regarding the risk of low back pain or injury [62].

### 2.3. Conclusions

Many different factors influencing the mechanical aspects of discomfort have been addressed using subject studies. The effects of different materials, the weight of the load, the region of discomfort, the duration of load carriage, and the center of mass have been investigated. In order to fully understand the factors responsible for discomfort during load carriage, the previous findings would have to be combined. However, this is often not possible because the types of investigated load carriage sys-
tems differ widely, as well as the population represented by the respective subject samples. A possible solution is to replace the subjects with a physical model, at the same time excluding the unwanted intra-subject and inter-subject variability. Of course, the value of such a physical model is tightly related to a solid model validation and there is currently no validated model available. This is probably due to a current lack of the necessary in vivo data for a validation and also a lack of appropriate sensors to measure such in vivo data.

Regarding the spinal loading during load carriage and the associated risk of low back pain or injury, the compression and shear forces in L4-L5 have been identified as risk factors. Models must be used to predict these forces, because in vivo measurements of the lumbar forces are almost impossible due to ethical reasons. Realistic models need to include the effects of postural adaptations and muscular activity during load carriage, which suggests the use of numerical simulations. Current challenges in numerical modeling of load carriage concern the mechanical interaction between the load carriage systems and the human body, as well as the unknown relevance of precise lumbar curvature data in spinal models.

2.4. References


33. Wang CXG, Chow DHK, Pope MH, editors. Biomechanical effect of load carriage on spine curvature and repositioning ability in adolescents. Proceedings of


3. Reliability of Pressure Measurements with Tekscan Sensors

Patrick D. Wettenschwiler¹,², Rolf Stämpfli¹, Silvio Lorenzetti², Stephen J. Ferguson², René M. Rossi¹, Simon Annaheim¹

Affiliation:

¹ Empa, Swiss Federal Laboratories for Materials Science and Technology, Lerchenfeldstrasse 5, CH-9014 St. Gallen, Switzerland
² Institute for Biomechanics, ETH Zurich, HCI E 355.1, CH-8093 Zürich, Switzerland


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3.1. Abstract

74% of adolescent backpack wearers suffer from neck or back pain. For other load carriage systems like protective vests and load bearing vests, overall musculoskeletal pain is significantly associated with carriage time. One factor, responsible for this kind of discomfort or pain, is the pressure at the body surface induced by the load carriage systems. Measurement of the pressure distribution at the interface between body and system is thus very important, but is currently still a technical challenge. Therefore, a subject study was conducted to determine the reliability of pressure distribution measurements on the skin of subjects wearing a protective vest and a load bearing vest. Measurements were taken with the subjects standing still and walking at 1.25m/s. Foil sensors were placed in four different areas of the upper body. The intra-subject test-retest reliability of the analysed average and peak pressures was low, but an extensive evaluation of different data processing steps increased the reliability. A comparison of the results revealed that the choice of sensor location and data processing may well be the key to achieving acceptable reliability in future measurements. However, measuring the pressure distribution on the body surface of human subjects wearing load carriage systems remains a challenge. Nevertheless, all mechanical interactions between body and environment may involve critical body surface pressure. Therefore, future improvement of the measurement reliability is crucial for a wide variety of industrial, medical and daily routine applications.

Keywords: body surface pressure; pressure sensor; reliability; data processing; artefact.

Abbreviations:

TE: typical error
CV: coefficient of variation
\( \text{av}_p \): average pressure over time
\( \text{av}_p \text{median} \): median pressure over time
\( \text{peak}_p \text{max} \): maximum pressure over time
\( \text{peak}_p \text{80} \): average of all local maxima equal to or above the 80\(^{th}\) percentile
3.2. Introduction

All mechanical contacts between the human body and its environment evoke body surface pressure. Even daily routine situations, such as sitting on a chair or carrying a backpack, may exceed critical pressure levels, inducing discomfort and pain. 74% of adolescent backpack wearers suffer from neck or back pain, validated by significantly poorer general health, more limited physical functioning, and more bodily pain [1]. Regarding other load carriage systems, such as load bearing vests, or ballistic vests, Konitzer et al. [2] found a significant association between wearing time and overall musculoskeletal pain (p=0.05). While several factors may be responsible for discomfort and pain, the pressure distribution (e.g. average or peak pressure values) on the body surface is certainly one of them: According to Stevenson et al. [3] the discomfort in the region of the shoulder during a 6km march depends almost one third on the average pressure measured on the shoulder contact area. In addition, focusing on the absolute values of the pressure distribution, the same study showed that 90% of the soldiers reported moderate discomfort with an average surface pressure of 20kPa. This is already above the value of 16kPa, at which Holloway et al. [4] showed that the skin blood flow decreases to essentially zero. However, maximum pressure levels during load carriage can be much higher: Hadid et al. [5] characterized the mechanical loading conditions in the shoulder region with a 25kg backpack and reported that the pressure on the skin reached values of up to 90kPa. This, in turn, shines a light on the occurrence of an uncommon yet serious injury, the impairment of the brachial plexus [6-8]. A retrospective study [9] revealed that in 100'000 Finnish soldiers, 53.7 soldiers per year (95% confidence interval: 39.5 - 67.8) suffered from impairment of the brachial plexus. The affected brachial plexus heals without special treatment, but the subject has to avoid any further exposure to backpacks or other load carriage systems until full recovery.

The measurement of the pressure distribution, and thus a possible optimization of it, is therefore important for a variety of medical issues, ranging from moderate discomfort to serious injury. Apart from the medical point of view, minimizing pain or discomfort is also in the interest of the manufacturer, as it is known to have considerable influence on user acceptance of a load carriage system [10, 11].
In previous studies assessing the pressure distribution on the skin of human subjects [12-14], Tekscan sensors (Tekscan, South Boston, MA, USA) were used. The biggest challenges with these sensors are the validity and reliability [15-17]. Measurement errors with Tekscan sensors can be as high as 34% [15] even for measurements conducted on flat surfaces. Little is known about the validity and reliability of measurements on the skin of human subjects, which can be much more complex regarding temperature, humidity, compliance and curvature of the surface. Therefore, the aim of this work was to investigate the intra-subject, test-retest reliability of Tekscan type 9801 sensor measuring the pressure distribution on the skin of human subjects wearing load carriage systems and whether the reliability is influenced by different factors, such as the activity of the subjects, the type of load carriage system or the location of the sensors.

Objectives:

1. What is the intra-subject, test-retest reliability of Tekscan type 9801 sensor measuring the pressure distribution on the skin of subjects wearing a ballistic vest (10.3kg) or load bearing vest (4.9kg)?

2. What is the influence of the activity of the subjects (standing still or walking at 1.25m/s) and the type of load carriage system (ballistic vest or load bearing vest) on the intra-subject, test-retest reliability of the pressure measurement?

3. Which areas of the subjects' body provide the highest intra-subject, test-retest reliability?

3.3. Materials and Methods

The study was approved by the Ethical Committee of the Canton of St. Gallen and was carried out in accordance with “The Code of Ethics of the World Medical Association” (Declaration of Helsinki, amended October 2013).

3.3.1. Participants

Eleven men with the following anthropometrical characteristics (mean ± standard deviation) were tested: age 24.9 ± 3.1 years, weight 80.6 ± 9.6kg, height 185.5 ± 7.7cm.
3.3.2. Load carriage systems

Two different systems were investigated (see Fig. 1): a ballistic vest weighing 10.3kg and a load bearing vest weighing 4.9kg, which are both currently in use in the Swiss Army. Underneath, the subjects wore the T-Shirt 90 and the Camouflage Suit 90/06 CO/PES of the Swiss Army.

Fig. 1. Load carriage systems used during the subject tests. Ballistic vest (left) and load bearing vest (right).

3.3.3. Pressure measurement and body regions

The pressure distribution was recorded using Tekscan type 9801 pressure sensitive foils (Tekscan, South Boston, MA, USA) with a pressure range up to 241kPa, according to the manufacturer. To enable optimal alignment of the sensors to the partly uneven surfaces of the human body, the sensors were cut along the designated lines (see Fig. 2). Every sensor was conditioned, equilibrated and calibrated prior to use under controlled conditions of 20 ± 2°C and 20 ± 5% relative humidity. The needed pressure was generated by steel blocks (flatness <0.1mm) and an underlying neoprene mattress of 4.5 ± 0.1mm thickness to enable a more homogenous distribution of pressure. For the conditioning, each sensor was loaded and unloaded at least 5 times in 3 second intervals with a pressure of 11.6 ± 0.1kPa. Equilibration was conducted using the algorithm provided by the Tekscan software under a pressure of 3.8 ± 0.1kPa. For the linear two-point calibration (0kPa is always included), 11.6 ± 0.1kPa were applied over an area of 75cm². Both the equilibration and the calibration were conducted, with a delay of 30 seconds between the placement of the weight.
and the recording to allow vibrations caused by the placing of the steel blocks to subside.

![Tekscan sensor 9801](image)

**Fig. 2. Tekscan sensor 9801.** 20 slits of 4.5cm length were cut into the sensor along the designated lines (bold lines). Image adapted from Tekscan product literature.

The sensors were placed directly onto the skin of the subjects in specific areas of the shoulder, chest, upper back and hip. As Jones and Hooper [18] observed at the shoulder backpack interface, measuring above or below the clothes made no significant difference (p=>0.05). As the sensors were non-adhesive, a skin-compatible spray (Leukospray, 09190-00, BSN medical, Hamburg, Germany) was applied. The exact locations of the sensors were defined in relation to specific anatomical landmarks as described in Table 1 and marked directly on the skin, using a cosmetic pen. Due to the natural variation within the somatotypes of the subjects, the placement of the sensors according to anatomical landmarks can be demanding. To minimize variation in the sensor placement and reduce possible effects on the measurements, this process was always conducted by the same person. The sensor located in the hip region was only considered while wearing the load bearing vest, as the ballistic vest did not cover this area. A sketch view of the sensor placement is provided in Fig. 3.

**Table 1. Definition of sensor placement according to anatomical landmarks.**

<table>
<thead>
<tr>
<th>Location</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulder</td>
<td>Most prominent point in the medial bend of the clavicle lies in the sensor area, 2cm from the right edge (when facing the subject) and 2cm from the bottom edge of the sensor area. Highest line between shoulder and neck lies perpendicular to the longitudinal edge of the sensor area.</td>
</tr>
<tr>
<td>Back</td>
<td>Lower corner of the scapula lies in the sensor area, 2cm from the bottom edge and 2cm from the left edge (when facing the back of the subject) of the sensor area. Spinous processes lies in the upper left corner (when facing the back of the subject) of the sensor area.</td>
</tr>
</tbody>
</table>
Chest | Middle of the sternum lies in the sensor area, 2cm from the upper right and 2cm from the upper left edge of the sensor area (when facing the subject). Nipple lies in the middle of the sensor area in lateral direction.

Hip | Sensor lies vertical and as lateral as possible with one quarter of the sensor area above and three quarters below the highest point of the iliac crest.

Fig. 3. Locations of the pressure sensitive foils.

Measurements were taken with two different load carriage systems and under two different activity conditions: standing still and walking on a treadmill at a speed of 1.25m/s. This speed lies within the 95% confidence interval (1.21 – 1.47m/s) of normal gait speed for men between 20 and 29, reported in a meta-analysis by Bohannon et al. [19]. To test the intra-subject, test-retest reliability, this series of measurements was conducted three times with each subject. All sensors were removed after each series of measurements and attached anew for the following test to minimize the effects of a possible misplacement.

To give the subjects time to get used to walking on a treadmill and to enable the sensors to adjust to skin temperature, the subjects walked for 2 minutes on the treadmill at a speed of 1.25m/s after each new placement of the sensors. Preliminary tests showed that the sensors adapt to skin temperature after a few seconds. However, additional preliminary tests indicated issues with humidity that penetrated the cells of the sensor if the subject was sweating considerably. Tekscan sensors measure compression via the decrease of electrical resistance in each sensor cell and as soon as
water soaked into a cell, the resistance decreased and the measurement led to an overestimation of pressure. As a consequence, the study was conducted in a climate chamber under controlled conditions of 20 ± 2°C and 20 ± 5% relative humidity. Additionally, the test protocol was designed to keep physical activity as low as possible.

In order to account for pressure artefacts due to bending of the sensor, base value measurements were recorded with the subjects standing with mounted sensors and fully dressed, yet unloaded by the vests. These base value measurements recorded a possible sensor output before donning a load carriage system, resulting from the bending of the sensors due to the curved body surface. Afterwards, the processed base value signals could be subtracted from the actual measurements (the measurements with load carriage systems) to minimize the artefact. We assumed linearity of the measurement based on the linear calibration of the sensors. With this, pressure resulting from the clothes could be compensated as well. These base value measurements were recorded repeatedly, so that all actual measurements were assigned a base value measurement that was conducted only shortly prior to the actual measurement.

All pressure measurements were recorded at 50Hz over a period of 6 seconds. The time period was chosen to be long enough to ensure that the breathing cycle was completed at least once during the measurement. The step cycle had an influence on the pressure measurement as well, but at a walking speed of 1.25m/s, step frequency was 3-5 times higher than the breathing frequency. To start all measurements under walking condition at the same moment in the step cycle, the investigator manually triggered the start of the measurement synchronously with the subject’s right heel strike.

3.3.4. Data processing and Statistics

Data processing was conducted in MATLAB (R2011b, The MathWorks, Natick, MA, USA).

Average pressure and peak pressure were evaluated, but rather than looking at the absolute values of the measurements, we focused on the intra-subject, test-retest reliability as measured by the typical error (TE) as coefficient of variation (CV) and the change in the mean [20]. As the first results of average pressure and peak pressure
revealed very poor intra-subject, test-retest reliability, they were evaluated in different ways in order to improve the reliability by improving the data processing method.

In case of average pressure, in a first step raw data were processed with and without offset correction. In the version without offset correction, a single value was calculated for each frame separately, taking into account all cells with a signal larger than zero. To apply offset correction, an average value over all frames was calculated for each sensor cell in the base value measurements. These values for each sensor cell were then deducted from each corresponding cell in every frame of the actual measurement before calculating the offset-corrected average pressure values in each frame. Each normal and each offset-corrected measurement thus produced a timeline of average pressure values. In a second step, a single value was computed from the timelines in two different ways, once by taking the average over time (av_paverage) and once by taking the median (av_pmedian).

In the case of peak pressure, raw data were also processed with and without offset correction, but an additional approach was made by filtering the raw data before offset correction with a rotationally symmetric Gaussian low pass filter of size 3x3 and standard deviation 0.5. This only made sense in the case of peak pressure. Average pressure would have been greatly altered by filtering the raw data, as all the cells that measured no pressure were excluded from the calculation of average pressure in each frame. Filtering would have included many of these cells by assigning them a value slightly larger than zero and thus lowering the average pressure of all non-zero cells. As this was not the case for peak pressure, three different versions of peak pressure timelines resulted from the first step of data processing. For these timelines, we also investigated two different ways of generating a single value. Once by simply taking the maximum (peak_pmax), and once by smoothing the timeline first with a moving average filter (see Fig. 4) and then averaging the values of all local maxima that were equal to or above the 80th percentile of the unfiltered timeline (peak_p80). The span of the moving average filter was set to 9 and a minimal distance of 20 (0.4 seconds) between two local maxima was defined. This version was applied to de-emphasize the effect of a possible single large peak value skewing the average of the remaining peak values during one measurement. Instead of reading out the highest single value, this method averaged the smoothed local maxima that were neither untypically low, nor untypically close together.
All applied steps of processing can thus be summarized as follows:

**Average pressure**

1. raw data were processed with and without offset correction
2. resulting timelines were further processed by calculating the average ($av_{p_{average}}$) as well as by extracting the median ($av_{p_{median}}$)

**Peak pressure**

1. raw data were processed with and without offset correction and with Gauss filtering plus offset correction
2. resulting timelines were further processed by taking the absolute maximum ($peak_{p_{max}}$) and by taking the average of all local maxima equal to or above the $80^{th}$ percentile of the unfiltered timeline ($peak_{p_{80}}$)

![Peak pressure timeline](image)

**Fig. 4. Peak pressure timeline.** Example resulting from pressure measurement on the hip with load bearing vest during activity condition walking. For the filtered signal, a moving average filter with a span of 9 was applied.

Before comparing the resulting reliabilities of all those different versions of processing, we checked the data for homoscedasticity. Peak pressure data clearly revealed heteroscedasticity, but the calculation of the TE as CV and the change in the mean is based on the assumption of homoscedasticity. In order to avoid biased values of reliability, we performed a $100\times\log$ transformation as suggested by Hopkins [20] and achieved the needed homoscedasticity. All data were therefore treated with
a 100*log transformation as the last step of data processing before calculating the reliability. Even though this was only necessary for peak pressure data, it was performed for the average pressure data as well for consistency reasons. The number of valid sets of measurements is given as n (Table 2-Table 6). A set consisted of three values from experiments under the same conditions, one from each series of measurements. If one or more of the values was zero, the whole set was excluded. This was necessary as the 100*log transformation was not applicable to a value of zero. The low values of n in peak\_p\textsubscript{80} in Table 3 are due to a number of measurements where no peaks in the smoothed timeline reached the 80\textsuperscript{th} percentile of the unfiltered timeline.

To compare the reliability of the average pressure according to load carriage systems and activities as shown in Table 4, we used av\_p\textsubscript{average} with offset correction, because it provided the best reliability of all evaluated versions of data processing. The same comparison was made for the reliability of the peak pressure (see Table 5), here we used peak\_p\textsubscript{max} with offset correction as it provided the best reliability.

To investigate the importance of the sensor location choice, the same data were also segregated by sensor location. It was not possible to segregate the data of an individual sensor location further by system or activity, as this would not have left enough values to calculate a meaningful reliability.

### 3.4. Results

The absolute values of the average pressure were in the range of 5 – 10kPa and those of the peak pressure in the range of 10 – 40kPa. Table 2 and Table 3 show the reliabilities of the average pressure and peak pressure, respectively, which evolved from different versions of data processing.
Table 2. Comparison of intra-subject, test-retest reliability in average pressure, achieved by different versions of data processing using typical errors (TE) as coefficients of variation (CV) and change in the mean. 2.-1. and 3.-2. refer to the comparison between the first and second and the second and third series of measurements, respectively. Superscript letters mark pairs with significant differences, based on non-overlapping 95% confidence intervals.

<table>
<thead>
<tr>
<th>Average pressure (100°log transf.)</th>
<th>av_p_average</th>
<th>av_p_median</th>
</tr>
</thead>
<tbody>
<tr>
<td>(average of average pressure timeline)</td>
<td>(median of average pressure timeline)</td>
<td></td>
</tr>
<tr>
<td>TE as CV [%]</td>
<td>Change in the mean [%]</td>
<td>n</td>
</tr>
<tr>
<td>2.-1.</td>
<td>3.-2.</td>
<td>2.-1.</td>
</tr>
<tr>
<td>No correction</td>
<td>53</td>
<td>55</td>
</tr>
<tr>
<td>Offset correction</td>
<td>50 f</td>
<td>55</td>
</tr>
</tbody>
</table>

Table 3. Comparison of intra-subject, test-retest reliability in peak pressure, achieved by different versions of data processing using typical errors (TE) as coefficients of variation (CV) and change in the mean. Superscript letters mark pairs with significant differences, based on non-overlapping 95% confidence intervals.

<table>
<thead>
<tr>
<th>Peak pressure (100°log transf.)</th>
<th>peak_p_max</th>
<th>peak_p80</th>
</tr>
</thead>
<tbody>
<tr>
<td>(maximum of peak pressure timeline)</td>
<td>(average peak height ≥ 80th percentile of unfiltered peak pressure timeline)</td>
<td></td>
</tr>
<tr>
<td>TE as CV [%]</td>
<td>Change in the mean [%]</td>
<td>n</td>
</tr>
<tr>
<td>2.-1.</td>
<td>3.-2.</td>
<td>2.-1.</td>
</tr>
<tr>
<td>No correction</td>
<td>91 a, b</td>
<td>86</td>
</tr>
<tr>
<td>Offset correction</td>
<td>60 a</td>
<td>73</td>
</tr>
<tr>
<td>Gauss filt., offset corr.</td>
<td>58 b</td>
<td>74</td>
</tr>
</tbody>
</table>

As can be seen in Table 2, calculating the average of the average pressure timeline showed better reliability than extracting the median. Furthermore, in case of the median, offset correction increased the CV and thus reduced reliability, but at the same time it clearly reduced the change in the mean for both median and average of the
average pressure timeline. According to Table 3, taking the maximum of the peak pressure timeline provided better results than taking the average peak height equal to or above the 80th percentile. Offset correction improved the reliability for both peak_pmax and peak_p80. Additional Gauss filtering before offset correction brought no improvement for peak_pmax and only slight improvement for peak_p80.

**Table 4.** Comparison of intra-subject, test-retest reliability in average pressure between load carriage systems and activities using typical error (TE) as coefficient of variation (CV) and change in the mean. Non-overlapping 95% confidence intervals did not occur when comparing the values of TE as CV and change in the mean across activity conditions or type of load carriage system.

<table>
<thead>
<tr>
<th>av_paverage (offset corr. &amp; 100*log transf.)</th>
<th>Ballistic vest</th>
<th>Load bearing vest</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>TE as CV [%]</td>
<td>Change in the mean [%]</td>
</tr>
<tr>
<td>2.-1. 3.-2.</td>
<td>2.-1. 3.-2.</td>
<td>2.-1. 3.-2.</td>
</tr>
<tr>
<td>Activity: stand</td>
<td>37 47</td>
<td>-6.4 -9.9</td>
</tr>
<tr>
<td>Activity: walk</td>
<td>42 43</td>
<td>-6.1 -1.5</td>
</tr>
</tbody>
</table>

**Table 5.** Comparison of intra-subject, test-retest reliability in peak pressure between load carriage systems and activities using typical error (TE) as coefficient of variation (CV) and change in the mean. Non-overlapping 95% confidence intervals did not occur when comparing the values of TE as CV and change in the mean across activity conditions or type of load carriage system.

<table>
<thead>
<tr>
<th>peak_pmax (offset corr. &amp; 100*log transf.)</th>
<th>Ballistic vest</th>
<th>Load bearing vest</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>TE as CV [%]</td>
<td>Change in the mean [%]</td>
</tr>
<tr>
<td>2.-1. 3.-2.</td>
<td>2.-1. 3.-2.</td>
<td>2.-1. 3.-2.</td>
</tr>
<tr>
<td>Activity: stand</td>
<td>61 70</td>
<td>1.2 -5.6</td>
</tr>
<tr>
<td>Activity: walk</td>
<td>74 74</td>
<td>-7.3 -4.8</td>
</tr>
</tbody>
</table>
Regarding the average pressure, activity influenced the CV in case of the load bearing vest, as static measurements (subjects standing still) led to better results than dynamic measurements (subjects walking) (Table 4). In the case of peak pressure, the static measurements provided better results for both vest types (Table 5). Peak pressure was generally less accurate than average pressure, but for both average and peak pressure the sensor location on the hip and the shoulder showed the best results, while reliability was worst in the sensor location on the back (Table 6).

Table 6. Comparison of intra-subject, test-retest reliability in average pressure and peak pressure between sensor locations using typical error (TE) as coefficient of variation (CV) and change in the mean. Superscript letters mark pairs with significant differences, based on non-overlapping 95% confidence intervals.

<table>
<thead>
<tr>
<th>Sensor locations</th>
<th>av_p_average (offset corr. &amp; 100*log transf.)</th>
<th>peak_pmax (offset corr. &amp; 100*log transf.)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>TE as CV [%]</td>
<td>Change in the mean [%]</td>
</tr>
<tr>
<td></td>
<td>2.-1. 3.-2.</td>
<td>2.-1. 3.-2.</td>
</tr>
<tr>
<td>Hip</td>
<td>24 a, b</td>
<td>-13.4 -5.9</td>
</tr>
<tr>
<td></td>
<td>25 a</td>
<td></td>
</tr>
<tr>
<td>Shoulder</td>
<td>28 c, d</td>
<td>-9.0 -11.5</td>
</tr>
<tr>
<td></td>
<td>44 l</td>
<td></td>
</tr>
<tr>
<td>Chest</td>
<td>57 a, c</td>
<td>6.3 8.4</td>
</tr>
<tr>
<td></td>
<td>37 g</td>
<td></td>
</tr>
<tr>
<td>Back</td>
<td>74 b, d</td>
<td>-19.9 -8.6</td>
</tr>
<tr>
<td></td>
<td>99 e, f, g</td>
<td></td>
</tr>
</tbody>
</table>

3.5. Discussion

3.5.1. Influence of data processing on intra-subject test-retest reliability

The resulting CVs are generally higher than expected. Measurement of similar yet less complex setups with Tekscan sensors in the past produced far less inaccuracy: Luo et al. [21] and Stevenson et al. [12] both found up to 20% measurement error; Brimacombe et al. [16] detected a maximal root mean square error of 24%; Hsiao et al. [15] measured up to 34% mean error. In contrast, our CVs were up to 55% for average pressure (Table 2) and up to 91% for peak pressure (Table 3) without further data processing. Alternative methods of calculating the average pressure (extracting
the mean, see Table 2) and the peak pressure (taking the average of all peaks equal to or above the 80th percentile, see Table 3) did not provide the expected improvement. However, offset correction brought significant improvement by reducing the change in the mean and in the case of peak pressure, also by reducing the CVs by up to 31 percentage points. By subtracting the base value signals from the actual measurements, pressure artefacts due to bending of the sensors or humidity that penetrated the cells of the sensors, could therefore be reduced. This step of the data processing method can be regarded as a valuable tool in assessing pressure distribution on the skin of human subjects. As the load carriage systems and the sensors have been taken off and placed anew for the repeated measurements, a part of the variation in the results may also originate from unintended changes in the exact position of the sensors and load carriage systems.

3.5.2. Possible improvements

Conditioning, equilibration and calibration of the sensors were conducted under conditions that resemble those of the actual measurements as suggested by Luo et al. [21] and Morin et al. [22]. However, the large positive change in the mean for all values between the first and second series, which could be significantly reduced through offset correction, suggests that humidity (from the sweat of the subjects) entered the sensor cells and thus increased sensitivity (Table 2 and Table 3). For a future setup, the edges of the sensors would have to be sealed prior to use, to prevent humidity from entering. An alternative would be to place a thin impermeable foil between skin and sensor. Once the offset correction was applied, this problem was accounted for, but then it became evident that most values of the change in the mean were slightly negative. This means that the pressure values tended to decrease from one series of measurements to the next. A change in the behaviour of the human soft tissue to this extent can be excluded, which leads to the assumption that the loss in sensitivity of the sensors already occurred during the measurements with one subject. This assumption is supported by the findings of Rose et al. [23], who detected a 21% decrease in plantar foot pressure after 72 gait cycles with Tekscan sensors. Komi et al. [24] also reported that the accuracy of an individual sensor decreased with repeated use. A study by Jansson et al. [25] recently pointed out that humidity is responsible for the diminishing load output of a Tekscan sensor. However, as Jansson et al. [25] conducted their first measurements in already wet condition, they did
not observe the initial increase in sensitivity that we experienced in our study. Overall, the evaluation of the change in the mean implies that repeated calibration or protection against humidity would be necessary between the individual repeated measurements with subjects. Another issue that needs to be discussed is the pressure range of the sensor. Due to the 8-bit resolution of the Tekscan sensors, a sensor with a pressure range of up to 240kPa has a maximal possible measurement resolution of 0.9kPa. When calibrating a sensor that has already been used one or more days earlier with a different subject, the pressure range of the sensor increased to more than twice the original value (e.g. to 621kPa). As the pressure range grew larger, the maximal possible measurement accuracy decreased. An improvement would be accomplished by choosing a smaller pressure range of the sensors, but this would induce the problem of missing peak pressure values that exceed the pressure range. There is the possibility of overlaying two sensors with different pressure ranges to cover a wide range and still have a good resolution in the lower part of the pressure range, but the increased thickness and stiffness might alter the measurement considerably.

3.5.3. Influence of load carriage system type and activity

Observing the data segregated by type of vest and activity (Table 4 and Table 5), static experiments mostly provided higher reliability than the dynamic experiments for both average pressure and peak pressure. This difference was not significant, but this tendency underlines the difficulties with artefacts on the sensors due to uncontrolled bending during motion. In case of the average pressure, measurements with the ballistic vest proved to be more reliable than with the load bearing vest, which could be due to the higher weight of the ballistic vest. However, for the peak pressure, no obvious tendency can be observed. It is possible that the higher stiffness of the ballistic vest, compared to the load bearing vest, was responsible for a decrease of reliability in the peak pressure values.

3.5.4. Influence of sensor location

Sensor location had a large influence on the intra-subject, test-retest reliability of the measurements. Table 6 shows that the sensor region in the back was associated with the lowest quality of measurement, while sensors in the regions of hip, shoulder and chest allowed for comparably lower CVs. The sensor region in the back includes the edges of the scapula, which generates a critical situation: a rather thin layer of
soft tissue covers an often protruding bone structure and curved surfaces are known to be more problematic for Tekscan measurements [12]. A positive outcome is that the shoulder area produced relatively good values, despite the thin layer of soft tissue that covers bony structures in this area like the lateral end of the clavicle. Sangeorzan et al. [26] showed that bony parts may be more critical regarding the oxygen supply under surface pressure. Assessing the pressure distribution of load carriage systems would not make much sense if the shoulder region had to be left out, as many load carriage systems include shoulder straps. The same holds true for the hip area, which is essential in all systems that include a hip belt. Focusing on the average pressure suggests that if sensor locations with very poor reliability like the area of the upper back are avoided, acceptable results are more likely to be achieved. CVs as low as 24% (Table 6) are not yet low enough to stop searching for improvement. However, these values appear in a different light when taking into account the complex conditions of body surface measurements, compared to measurements on rigid flat surfaces as conducted by Luo et al. [21] and Hsiao et al. [15], which also yielded measurement errors up to 20%, and 34%, respectively.

3.5.5. Limitations

When load carriage systems are worn for long periods, fatigue may affect the pressure distribution. However, the focus of this study lies on the measurement of the momentary pressure distribution. Furthermore, the exclusion of the sets of measurements that included one or more zero values was necessary due to the logarithmic transformation, but this may have affected the resulting reliability. This exclusion is justified, as the measurements in this study are focused on the areas where possibly critical pressure occurs and not on areas with no or almost no pressure on the body surface.

3.5.6. Outlook

Reflecting on the extent of the CVs, human subjects must be regarded as a challenging and inconsistent model for the analysis of pressure distribution with Tekscan sensors on the body surface under load. This reveals the need for a standardised dummy to evaluate different load carriage systems. On a dummy, measurement reliability is expected to be considerably higher, as can be seen in experiments conducted by Stevenson et al. [12]. This would also solve the problems that originate from differ-
ences in the anatomy of the subjects and different gait patterns. Nevertheless, for the validation of such a dummy, measurements on human subjects are crucial, thus we need to optimize the reliability. As improvement by data processing is not sufficient, changes in the experimental setup might be necessary, namely the choice of optimal areas for sensor location and protection against sweat entering the sensor cells.

Future studies should also include repeated measurements without taking off the load carriage systems and sensors, in order to quantify how much variation is due to differences in the positioning of the sensors and load carriage systems.

3.6. Conclusions

A subject study was conducted to determine the reliability of pressure distribution measurements on the skin of subjects wearing a protective vest and a load bearing vest. Measurements were taken with the subjects standing still and walking at 1.25m/s. Foil sensors were placed in four different areas of the upper body. Intra-subject, test-retest reliability was generally much lower than expected, but data processing brought substantial improvement: Offset correction reduced the change in the mean significantly and in the case of peak pressure, it also significantly reduced the CVs by up to 31 percentage points. Further improvement is expected if the sensors will be protected against humidity entering the sensor cells in future measurements. Significant differences in reliability evolved when comparing the chosen sensor locations, where a thick layer of soft tissue covering the underlying bone structures proved to be of advantage. Measuring the pressure distribution on the body surface of human subjects wearing load carriage systems remains a challenge and in this type of measurements, promising reliability has not yet been reached with Tekscan sensors.

3.7. Acknowledgments

We thank armasuisse for their financial support. Apart from funding, armasuisse was not involved in this study. In addition, the authors gratefully acknowledge Veronika Meyer for proofreading and improvement of the publication.
3.8. References


4. Mechanical Predictors of Discomfort during Load Carriage

Patrick D. Wettenschwiler¹,², Silvio Lorenzetti², Rolf Stämpfli¹, René M. Rossi¹, Stephen J. Ferguson², Simon Annaheim¹

Affiliation:

¹ Empa, Swiss Federal Laboratories for Materials Science and Technology, St. Gallen, Switzerland

² Institute for Biomechanics, ETH Zurich, Zurich, Switzerland


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4.1. Abstract

Discomfort during load carriage is a major issue for activities using backpacks (e.g. infantry maneuvers, children carrying school supplies, or outdoor sports). It is currently unclear which mechanical parameters are responsible for subjectively perceived discomfort. The aim of this study was to identify objectively measured mechanical predictors of discomfort during load carriage. We compared twelve different configurations of a typical load carriage system, a commercially available backpack with a hip belt. The pressure distribution under the hip belt and the shoulder strap, as well as the tensile force in the strap and the relative motion of the backpack were measured. Multiple linear regression analyses were conducted to investigate possible predictors of discomfort. The results demonstrate that static peak pressure, or alternatively, static strap force is a significant (p<0.001) predictor of discomfort during load carriage in the shoulder and hip region, accounting for 85% or more of the variation in discomfort. As an additional finding, we discovered that the regression coefficients of these predictors are significantly smaller for the hip than for the shoulder region. As static peak pressure is measured directly on the body, it is less dependent on the type of load carriage system than static strap force. Therefore, static peak pressure is well suited as a generally applicable, objective mechanical parameter for the optimization of load carriage system design. Alternatively, when limited to load carriage systems of the type backpack with hip belt, static strap force is the most valuable predictor of discomfort. The regionally differing regression coefficients of both predictors imply that the hip region is significantly more tolerant than the shoulder region. In order to minimize discomfort, users should be encouraged to shift load from the shoulders to the hip region wherever possible, at the same time likely decreasing the risk of low back pain or injury.
4.2. Introduction

Discomfort during load carriage is a major issue for activities using backpacks (e.g. infantry maneuvers, children carrying school supplies, or outdoor sports). According to Sheir-Neiss et al. [1], 74% of adolescent backpack wearers suffer from neck or back pain, validated by significantly poorer general health, more limited physical functioning, and more bodily pain. As loads have increased during the last decades [2, 3], discomfort during load carriage will become an even more important topic. With increasing loads, discomfort is more likely to be accompanied by injuries like the impairment of the brachial plexus [4-8] or low back pain or injury [9-11]. Any efforts to reduce the occurrence of these medical issues, ranging from discomfort to severe injury, are thus most welcome. A potential improvement could be achieved by optimizing load carriage system design, e.g. system structure or the material properties at the interface between the system and the body. Partial success towards this goal has already been achieved through the use of load carriage systems with a hip belt or a comparable structure, supporting a load shift from the shoulders to the hip [12]. Manufacturers of load carriage systems have an additional motivation to improve their design, as discomfort is known to have considerable influence on user acceptance [13, 14]. However, the variety of currently available load carriage systems suggests that the optimum has not yet been reached. The optimization is hindered by the fact that discomfort is a subjective perception and also depends on the length of time the system is worn [15]. In addition, the personal mood of the subjects and their physical constitution may influence the subjective perception of discomfort [16, 17]. Therefore, objective measurement of discomfort during load carriage is currently very challenging.

There are several definitions of comfort and discomfort available in the literature [18]. According to Richards et al. [19], ‘Comfort is a continuous dimension of experience – varying from strongly positive (very comfortable) to strongly negative (very uncomfortable).’ Discomfort depends on biomechanical factors that are responsible for feelings of pain, soreness, numbness, stiffness and other comparable perceptions [20, 21]. Additionally, thermal aspects may influence discomfort, but they are well understood [22, 23]. In contrast, little is known about the mechanical aspects of discomfort: Local ischemia in the skin and underlying muscular and neural tissue is expected to induce discomfort: Holloway et al. [24] reported blood occlusion to occur at a body surface
pressure of 16kPa. Sangeorzan et al. [25] suggested even lower pressure values between 5.6kPa and 9.5kPa to be sufficient for ischemia. The direct compressive force of a backpack’s shoulder strap was shown to induce increased local fatigue in the upper trapezius muscle during an exhausting arm abduction task [26]. Regarding recommended limits for mechanical parameters, Bryant et al. [27] investigated a static scenario and proposed a maximal lumbar force of 135N and a maximal shoulder force of 145N for load carriage. The same study reported that an average pressure of 20kPa in the shoulder region resulted in discomfort for 90% of the soldiers examined. In light of these scarce findings in the literature, it is not yet fully understood which mechanical parameters are directly responsible for subjectively perceived discomfort. There are many open questions, e.g. which parameters have the largest influence on discomfort? Furthermore, is it required to differentiate between static and dynamic values of mechanical parameters like pressures and forces? It is evident that gaining a comprehensive knowledge of the mechanical predictors is a crucial step towards minimizing discomfort during load carriage. Therefore, the aim of this study was to identify objectively measured mechanical predictors of discomfort during load carriage, using pressure sensing mats, strap force sensors and a 3D motion tracking system.

4.3. Materials and Methods

Ten male subjects without any history of back pain and with the following anthropometrical characteristics were tested (mean ± standard deviation): age 28.0 ± 3.7 years, weight 73.1 ± 10.4kg, height 178.7 ± 5.7cm. All subjects provided written informed consent.

This study was approved by the Ethical Committee of the Canton of St. Gallen and was carried out in accordance with “The Code of Ethics of the World Medical Association” (Declaration of Helsinki, amended October 2013).

4.3.1. Load Carriage System

The load carriage system applied in this study is the commercially available backpack “Deuter ACT Lite 50+10” (Deuter Sport GmbH, Gersthofen, Germany). According to the manufacturer, it is intended for use in a wide variety of activities, including trekking, alpine tours and travelling. Two modifications were made to the system for this
study: Firstly, all metal parts had to be removed due to the use of electromagnetic sensors. Secondly, an efficient change of the payload, while keeping the center of mass constant in all three dimensions, had to be enabled.

4.3.2. Modifications

All modifications respected lateral symmetry of the backpack. We removed the aluminum rods, which built the frame in the back wall of the backpack. A wooden box (65.0cm height, 27.5cm length, 15.8cm depth, and 0.9cm thickness) was inserted into the backpack and fixed to the back wall of the backpack. The rigid connection between the backpack and the wooden box replaced the function of the aluminum rods, enabling load transfer between the hip belt and the shoulder straps. Two openings in the backpack and the wooden box were created for easy access to a modular payload, one at the bottom and one at the top. The payload was constructed out of cardboard boxes filled with sand and resulted in a steady center of mass, positioned 30.5cm from the bottom (Fig. 1).

Fig. 1. Modified load carriage system. The modular payload is shown on the left.
4.3.3. Configurations

We compared twelve different configurations of one typical load carriage system, thus eliminating potential effects of the design on discomfort. Mackie et al. [28] reported that load weight and hip belt use have the largest effects on interface pressure and strap forces. Therefore, we defined twelve configurations resulting from a combination of three different loads and four different hip belt lengths: The total masses of the load carriage system were 15.0kg, 20.0kg, and 25.0kg. The hip belt lengths were calibrated for each subject, corresponding to 30N, 60N, 90N, and 120N of tension during initial upright standing with the 20.0kg load. As the tension in the hip belt is sensitive to breathing and hip joint motion, hip belt length is considered to be more appropriate than hip belt tension to define the load carriage system configurations.

4.3.4. Experiments

For all measurements, subjects wore running shorts and sneakers, but no shirt. The back length of the load carriage system was adjusted for each subject to position the upper end of the hip belt level with the highest point of the iliac crest. The lengths of the hip belt were marked according to the predefined tension levels. In an acclimatization phase, the subjects walked with the load carriage system on the treadmill at 4.5km/h, until they felt at ease. One subject could not adjust to comfortable walking at this speed, and instead felt comfortable walking at 4.0km/h. This subject consequently performed all measurements at 4.0km/h. As soon as the subjects felt at ease, they were asked to stop and to fill out the discomfort questionnaire according to the currently perceived discomfort. These answers were not evaluated, as they served as a familiarization trial.

To minimize the effect of minor differences in the exact placement of the load carriage system, three iterations were performed to measure the mechanical parameters, with a break of 20 minutes in between. These iterations were identical, except for the order in which the load carriage system configurations were assessed: Firstly, as a warm-up, subjects walked on the treadmill for two minutes with a total load of 15.0kg. Then one measurement was performed without the load carriage system, to record possible artifacts in the pressure sensors due to bending. The reason for this measurement without load is explained in more detail in the section “Body Surface Pressure” below. Afterwards, all load carriage system configurations were applied in a
randomized order. All randomizations for our study were based on true random numbers provided by RANDOM.ORG [29]. For every configuration, a static measurement was followed by a dynamic measurement, each of them lasting for six seconds at a sampling rate of 120Hz. During the static measurement, subjects stood upright, facing straight ahead. During the dynamic measurement, subjects walked on the level treadmill. The dynamic measurement was triggered by the investigator simultaneously with the right heel strike of the subjects, as soon as they were walking regularly at the given speed.

After completing the measurements of the mechanical parameters, the subjects were asked to report their discomfort for each configuration in randomized order. For this discomfort assessment, the subjects walked on the treadmill for one minute with each configuration, before stopping and filling out the discomfort questionnaire regarding the currently perceived discomfort with the corresponding configuration. By assessing the discomfort at the very end of the experiments and choosing a small duration for the last treadmill walk with each configuration, we prevented substantial differences in total prior exposition time between the configurations.

4.3.5. Measured Parameters

4.3.5.1. Body Surface Pressure

Body surface pressure was recorded using Tekscan type 9811E pressure sensitive foils (Tekscan, South Boston, MA, USA) with a pressure range up to 172kPa, according to the manufacturer’s specifications. The sensitive foil comprises 6 x 16 sensor cells, covering an area of 7.6cm x 20.3cm. Prior to use, the sensors were conditioned, equilibrated, and calibrated according to the manufacturer’s recommendations. Equilibration was performed at 20kPa, while two-point calibration was performed at 20kPa and 50kPa. The sensors were placed on the right shoulder and right hip region of the subjects, as shown in Fig. 2. The individual in Fig. 2 has given written informed consent (as outlined in PLOS consent form) to publish these pictures. For a precise placement of the pressure sensors according to anatomical landmarks and for easier handling, the subjects wore no shirt. The sensitivity of Tekscan sensors to humidity and temperature has been reported in a previous study [30]. To protect the sensors from humidity, the sensors were welded into 0.05mm thick polyethylene pro-
tective covers. As the clothing layers have no influence on the pressure [31], thin polyethylene layers are not expected to have an influence on the pressure. To minimize a possible change in temperature during the measurements, the sensors were worn during the two-minute warm-up walk. Newly calibrated sensors were used for every measurement iteration step.

Fig. 2. Subject with pressure sensors (left) and load carriage system (right). On the right side, black arrows point to the locations of the strap force sensors on the shoulder strap and on the hip belt and white arrows point to the locations of the Polhemus sensors on the shoulder and on the hip.

The recorded pressure distribution data was processed in MATLAB (R2012b, The MathWorks, Natick, MA, USA) to determine the average pressure and the peak pressure of each measurement. With a measurement duration of six seconds and a sampling rate of 120Hz, every measurement consisted of 720 frames. An offset correction was performed first to account for possible artifacts due to bending of the sensors: Using the measurements without load carriage system, a base average value was calculated over all frames for each cell of the sensors. The base values were subtracted from all corresponding measurements with the load carriage system. After
this offset correction, the average pressure and the peak pressure were calculated for static and dynamic measurements in the shoulder and the hip region. The average pressure value was calculated by first taking the mean of all non-zero cells in each measured frame. With these mean values of each frame, it was further possible to calculate the mean value over time to reach the average pressure of each measurement. Similarly, the peak pressure was calculated by first taking the maximum value in each frame. With these maximum values for each frame, it was further possible to extract the maximum across time to reach the peak pressure of each measurement. For each subject, the final average and peak pressures were calculated by taking the mean of all three iterations.

4.3.5.2. Strap Force

The forces were measured in the right shoulder strap and in the hip belt using force sensors based on strain gauges (Fig. 2). To determine the strap forces for each measurement, the mean value over time was calculated. For each subject, the strap forces were calculated by taking the mean of all three iterations.

4.3.5.3. Relative Motion

The 3D motion tracking system Polhemus Liberty (Polhemus, Colchester, VT, USA) was used to investigate the relative motion between the bulk of the load carriage system and the body. The electromagnetic field source of the tracking system was mounted at the posterior top of the load carriage system (Fig. 1). Using double-sided adhesive tape, sensors were mounted on the acromion of the right shoulder and on the hip fin of the load carriage system at the height of the underlying anterior superior iliac spine (Fig. 2). For each measurement, the cumulative change of distance between sensor and source in all three axes, divided by the measurement duration, was calculated for the relative motion value. For each subject, the final relative motion was calculated by taking the mean of all three iteration steps.

4.3.5.4. Discomfort

The focus of this study lies on the mechanical aspects of discomfort, therefore thermal aspects of discomfort were minimized as much as possible. The lab environment (air conditioning) was adjusted to the subjects’ individual preferences and the study design featured only moderate activity.
For the subjective discomfort scores in our study, the subjects were asked to rate the currently perceived discomfort on a visual analogue scale, ranging from “no discomfort”, corresponding to a value of 0, to “maximal discomfort”, corresponding to a value of 10. For every load carriage system configuration, one score for the shoulder region, one for the hip region, and one for the overall discomfort was recorded.

4.4. **Statistical Analysis**

For each parameter, the mean of all subjects was calculated for every load carriage system configuration in order to exclude the effect of the intra-subject variation in the perception of discomfort. Using these mean values of all subjects, several multiple linear regression analyses were conducted using IBM SPSS (22.0, IBM Corp., Armonk, NY, USA) to identify the mechanical predictors of discomfort (Table 1). In a first approach, only the pressure parameters (average pressure and peak pressure) were entered as independent variables. When using these parameters as possible predictors, the type and design of the load carriage system is not relevant. Therefore the results of these regressions are also valid for other types of load carriage systems. In a second round, all parameters were entered as independent variables. Consequently, these results are only valid for load carriage systems of the type backpack with hip belt. However, rather than conducting a regression analysis with eleven degrees of freedom and eight possible predictors, separate regressions were conducted for the static and dynamic parameters in the second round. All these regressions were calculated for the shoulder and hip region separately.
**Table 1.** List of independent variables included in multiple linear regressions. The dependent variable was regional discomfort.

<table>
<thead>
<tr>
<th></th>
<th>Regression with pressure parameters</th>
<th>Regression with all static parameters</th>
<th>Regression with all dynamic parameters</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Average pressure [kPa]</strong></td>
<td>static</td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td></td>
<td>dynamic</td>
<td>x</td>
<td></td>
</tr>
<tr>
<td><strong>Peak pressure [kPa]</strong></td>
<td>static</td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td></td>
<td>dynamic</td>
<td>x</td>
<td></td>
</tr>
<tr>
<td><strong>Strap force [kPa]</strong></td>
<td>static</td>
<td></td>
<td>x</td>
</tr>
<tr>
<td></td>
<td>dynamic</td>
<td></td>
<td>x</td>
</tr>
<tr>
<td><strong>Relative motion [kPa]</strong></td>
<td>static</td>
<td></td>
<td>x</td>
</tr>
<tr>
<td></td>
<td>dynamic</td>
<td></td>
<td>x</td>
</tr>
</tbody>
</table>

Where the regression results revealed the same single parameter as significant predictor for discomfort in both regions, a direct comparison of both regions is warranted. Hence, the linear relationship of these parameters between the shoulder and the hip region at equal discomfort was calculated from the corresponding regression equations.

Finally, one multiple linear regression was conducted with overall discomfort as dependent variable and with shoulder discomfort and hip discomfort as independent variables. All regression analyses in this study were conducted using backwards elimination method, applying Bonferroni correction for multiple testing. The significance level was defined at p<0.05.

**4.5. Results**

The absolute values of the discomfort and the objectively measured mechanical parameters are not the primary interest in this study. Nevertheless, they are provided as a mean of all subjects for all configurations as supporting information (S1-S3 Tables).

The results of the multiple linear regression analysis in both investigated regions are shown in Fig. 3, with discomfort as dependent variable and the pressure parameters as independent variables. The regression coefficients of static peak pressure exhibit
the following 95% confidence intervals: 0.038 - 0.067 for the hip and 0.079 - 0.144 for the shoulder region.

Fig. 3. Results of the regression analyses using average and peak pressure (static and dynamic) as independent variables. Multiple linear regression analyses revealed static peak pressure as significant (p<0.001) predictor of discomfort in the shoulder and hip region. The non-significant predictors were removed from the model during the backwards elimination. Data points show the subject's mean ± standard error of measurement for each configuration, dotted lines show 95% prediction intervals. Regression equations: \( y = 0.111x + 2.102, R^2 = 0.85 \) (shoulder); \( y = 0.052x + 1.127, R^2 = 0.86 \) (hip).

The results of the multiple linear regression analyses with all mechanical parameters as independent variables are presented separately for the static and dynamic parameters in Table 2.
Table 2. Results of the regression analyses using average pressure, peak pressure, strap force, and relative motion as independent variables. The dependent variable was regional discomfort. Only the variables with significant coefficients are listed in the table, the other variables were removed from the model during the backwards elimination. * p<0.001, ** p<0.01

<table>
<thead>
<tr>
<th>Condition</th>
<th>Region</th>
<th>$R^2$</th>
<th>Model</th>
<th>Coefficients</th>
<th>95% confidence interval of coefficients</th>
<th>Standardized coeff.</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>lower</td>
<td>upper</td>
</tr>
<tr>
<td>Static</td>
<td>Shoulder</td>
<td>0.91</td>
<td>Constant</td>
<td>-1.339</td>
<td>-2.590</td>
<td>-0.088</td>
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<tr>
<td></td>
<td>Hip</td>
<td>0.85</td>
<td>Constant</td>
<td>0.884</td>
<td>-0.015</td>
<td>1.782</td>
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<tr>
<td>Dynamic</td>
<td>Shoulder</td>
<td>0.96</td>
<td>Constant</td>
<td>6.984</td>
<td>2.371</td>
<td>11.596</td>
</tr>
<tr>
<td></td>
<td>Hip</td>
<td>0.94</td>
<td>Constant</td>
<td>-11.275</td>
<td>-19.502</td>
<td>-3.049</td>
</tr>
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<td></td>
<td></td>
<td></td>
<td>Strap force</td>
<td>0.097</td>
<td>0.076</td>
<td>0.118</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.040</td>
<td>0.029</td>
<td>0.052</td>
<td>0.922*</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>-0.836</td>
<td>-1.291</td>
<td>-0.382</td>
<td>-0.375**</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.041</td>
<td>0.032</td>
<td>0.050</td>
<td>0.909*</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>1.049</td>
<td>0.345</td>
<td>1.752</td>
<td>0.286**</td>
</tr>
</tbody>
</table>

For a direct comparison between both regions, the linear relationship between static peak pressure in the shoulder ($p_{shoulder}$) and in the hip region ($p_{hip}$) at equal discomfort in both regions is given in Equation (1). The linear relationship between static strap force in the shoulder ($f_{shoulder}$) and in the hip region ($f_{hip}$) at equal discomfort in both regions is given in Equation (2).

$$p_{hip} = 2.135 p_{shoulder} + 18.750 \quad (1)$$

$$f_{hip} = 2.425 f_{shoulder} - 55.575 \quad (2)$$

The results of the multiple linear regression analysis with overall discomfort as dependent variable and regional discomfort as independent variables are shown in Table 3.
Table 3. Results of the regression analysis using shoulder discomfort and hip discomfort as independent variables. The dependent variable was overall discomfort. $R^2 = 0.90$, *$p<0.001$

<table>
<thead>
<tr>
<th>Model</th>
<th>Coefficients</th>
<th>95% confidence interval of coefficients</th>
<th>Standardized coefficients</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>lower</td>
<td>upper</td>
</tr>
<tr>
<td>Constant</td>
<td>-0.501</td>
<td>-1.671</td>
<td>0.652</td>
</tr>
<tr>
<td>Shoulder discomfort</td>
<td>0.967</td>
<td>0.476</td>
<td>0.865</td>
</tr>
<tr>
<td>Hip discomfort</td>
<td>0.463</td>
<td>0.292</td>
<td>0.634</td>
</tr>
</tbody>
</table>

4.6. Discussion

The aim of this study was to identify objectively measured mechanical predictors of discomfort during load carriage. We compared twelve configurations of a typical load carriage system, evaluating average pressure, peak pressure, strap forces and relative motion between the bulk of the system and the body in the shoulder and hip region for static and dynamic conditions. We conducted the main multiple linear regression analyses twice: once using only the pressure parameters as independent variables and once using all mechanical parameters as independent variables. For both versions, we used regional discomfort as dependent variables.

4.6.1. Static Peak Pressure as Predictor of Discomfort

Most of the static peak pressure values found in this study are higher than 16kPa, at which Holloway et al. [24] reported blood occlusion to occur (Fig. 3). Consequently, the presence of discomfort in this study is consistent with expectations.

Our results show that static peak pressure, as significant predictor of discomfort, accounts for 85% of discomfort in the shoulder region and for 86% of discomfort in the hip region (Fig. 3). These results indicate that mechanical parameters are more powerful predictors of discomfort than literature has previously suggested. In a similar study, Stevenson et al. [32] found average pressure to be the best predictor of discomfort, accounting for 31% of its variation. Compared to the results found in our study, this value reported in literature is lower, which could be due to differences in
the design. While the data of Stevenson et al. [32] originated from a comparison of nine different load carriage systems, our study compared twelve different configurations of one load carriage system. Additionally, Stevenson et al. [32] measured the mechanical parameters on a human load carriage simulator, while our measurements were conducted on subjects. Both designs are justified by their respective advantages. A human simulator usually provides better repeatability, whereas human subjects are more realistic. The use of different load carriage systems provides a broader generalization of the findings, whereas by using only one system, as in this study, several otherwise uncontrolled parameters are kept constant. At the same time, we are fully aware that the type of the investigated load carriage system may influence the outcome of a study. However, by using only the pressure parameters for these regressions, we strove to minimize dependence of the measurements on the load carriage system type. The perception of static peak pressure on the body surface does not depend on the type of load carriage system used. Hence, static peak pressure is well suited as generally applicable, objective mechanical parameter for the optimization of load carriage system design. Such an objective mechanical parameter is of great value for the improvement of load carriage system design, so that in the future, users may profit from decreased discomfort during load carriage.

4.6.2. Strap Force and Relative Motion as Predictors of Discomfort

To investigate whether the mechanical parameters that were measured on the load carriage system could explain additional variance in discomfort, we further conducted multiple linear regression analyses with all mechanical parameters as independent variables.

In static conditions, strap force is a significant (p<0.001) predictor of discomfort in the shoulder and the hip region (Table 2). Therefore, static peak pressure is not the best predictor of discomfort, if static strap force data is considered simultaneously. Static peak pressure does not account for any significant additional variance in discomfort either and is excluded from the models. The explanation for these results lies in the presence of high multicollinearity; among the parameters static average pressure, dynamic average pressure, static peak pressure, dynamic peak pressure, static strap force, and dynamic strap force, the Pearson correlation coefficient was above 0.9 for all comparisons. All regression results contain only one of these parameters. As a
consequence, in the absence of static peak pressure or strap force values, other parameters of this group could also be significant predictors of discomfort. Our regression results are not coincidental, as the applied backwards elimination method eliminates the least significant predictor in each step, until all of the remaining predictors meet the predefined significance criterion. One must keep in mind that excluded parameters, like static average pressure, dynamic average pressure, and dynamic peak pressure, are not necessarily invalid as predictors of discomfort. In our sample, they are simply less powerful than static peak pressure or strap force, which we found to be the best predictors of discomfort.

In dynamic conditions, strap force ($p<0.001$) and relative motion ($p<0.01$) between the bulk of the load carriage system and the body are significant predictors of discomfort (Table 2). For both regions, relative motion is a less important predictor of discomfort than strap force, as can be seen by the standardized coefficients (Table 2). While more relative motion is associated with more discomfort for the hip region, the opposite was found for the shoulder region. More relative motion in the shoulder region was therefore associated with less discomfort. A possible explanation for this unexpected finding can be found in the results of Sharpe et al. [33]: Compared to the natural walking pattern without load, a backpack reduces the relative phase of the rotation between pelvis and thorax [33]. An increasing relative motion in the shoulder region may therefore be a sign of less restriction of the shoulder girdle through the load carriage system, enabling a more natural motion pattern and resulting in less discomfort. Nevertheless, despite explaining 96% (shoulder) and 94% (hip) of variation in discomfort, our models with dynamic strap force and dynamic relative motion as predictors have to be treated with care. In static conditions, in contrast, the models with strap force as sole significant predictor are conclusive. A big advantage of strap forces is that they are usually much easier to measure than static peak pressures. While these results should not be generalized to any possible type of load carriage system, we consider them to be valid for systems that are comparable to the type used in this study, i.e. backpacks with hip belts. Thus static strap forces can be regarded as valuable predictors of discomfort in both regions when applied to load carriage systems of the type backpack with hip belt.
4.6.3. Regional Differences in the Perception of Discomfort

Aside from providing objective mechanical parameters to predict discomfort during load carriage, our results also offer some valuable information about optimal adjustments of load carriage systems.

Due to the non-overlapping confidence intervals of the regression coefficients of static peak pressure in the shoulder and the hip region, we can deduce that the hip region is significantly more tolerant than the shoulder region, regarding an increase in static peak pressure. The same is true for static strap force (Table 2). According to the quantitative comparison in Equation (1), static peak pressure in the hip needs to be more than twice the static peak pressure in the shoulder to cause the same amount of discomfort. These results are in line with the findings of Scribano et al. [34], who reported the hip region to be two to three times more tolerant, regarding the absolute values of local skin pressure. More recently, Martin et al. [35] suggested the hip region to be less sensitive than the shoulder region to an increase in pressure.

We were able to confirm the finding of Martin et al. (see Equation (1)). The hip region is also less sensitive than the shoulder region to an increase in static strap force (see Equation (2)). In order to minimize discomfort, users should therefore be encouraged to shift load from the shoulders to the hip region wherever possible. This is accompanied by another benefit, because as more load is applied at the hip, less load impacts on the spine. Hence, shifting load to the hip at the same time likely decreases the risk of low back pain or injury.

4.6.4. Overall Discomfort

As this study focused on discomfort in the shoulder and hip region only, it was important to analyze their association with overall discomfort. Shoulder discomfort and hip discomfort are both significant (p<0.001) predictors of overall discomfort (Table 3). Together, they can explain 90% of the variation in overall discomfort. From these results, we deduce that for this study, focusing on the shoulder and hip region was adequate to draw conclusions also on overall discomfort.

4.6.5. Limitations

The subjects used in this study were all male and young (28.0 ± 3.7 years). It is not clear to which extent our findings apply also to females and/or older people. Potential
factors of variability within the sample of young male subjects include body composition and regional subcutaneous fat distribution. Their influence remains to be investigated in future studies. In addition, the pressure measurements conducted in this study only recorded pressure normal to the body surface. Tangential stress and resulting shear strain in the skin and the underlying tissue is not included in the pressure parameters. However, other parameters, e.g. strap forces, do not differentiate between vertical and tangential pressure and thus shear strain is indirectly considered in our investigations. The most challenging issue was a possible temperature difference of the pressure sensors between calibration and measurement on the skin. To minimize this, sensors were given enough time to adjust to skin temperature and the order of configurations tested was randomized. This enabled an unbiased comparison between configurations using the mean of all subjects. Regarding the load carriage system used in our study, the necessary modifications may have influenced the discomfort perceived by the subjects. Therefore, the discomfort values reported in this study cannot be applied to the commercially available load carriage system in its original state. Finally, instead of letting the subjects walk at a self-selected walking speed, we asked the subjects to walk at 4.5km/h, as a uniform walking speed naturally occurs in typical fields of load carriage, e.g. trekking/hiking in a group, infantry. One subject could not adjust to comfortable walking at 4.5km/h. Instead of enforcing an uncomfortable and possibly unnatural walking pattern, we measured this subject at a walking speed of 4.0km/h. The statistical power of this study is affected by the assumption of linear relationships between discomfort and the predictors, the use of mean values of all subjects, and the confined number of ten subjects.

4.7. Conclusions

In this study, we found that static peak pressure, or alternatively, static strap force, is a significant (p<0.001) predictor of discomfort during load carriage in the shoulder and hip region, accounting for 85% or more of the variation in perceived discomfort. As an additional finding, we discovered that the regression coefficients of these predictors are significantly smaller for the hip than for the shoulder region. For a quantitative comparison, the linear relationship of static peak pressure and static strap force between both regions at equal discomfort was evaluated. This revealed the shoulder region to be more than twice as sensitive as the hip region to an increase in static peak pressure or static strap force. In order to minimize discomfort, users should be
encouraged to shift load from the shoulders to the hip region wherever possible, at the same time likely decreasing the risk of low back pain or injury. However, the main outcome of this study is the successful identification of objectively measured mechanical predictors of discomfort, which represent a valuable tool for the optimization of load carriage system design.

4.8. Acknowledgments

We would like to thank Alexander Haag for the laser-welding of protective bags for the pressure sensors.

4.9. References


26. Piscione J, Gamet D. Effect of mechanical compression due to load carrying on shoulder muscle fatigue during sustained isometric arm abduction: an


### 4.10. Supporting Information

**S1 Table. Discomfort scores.** For each measured configuration, the mean of all subjects ± the standard error of measurement is shown.

<table>
<thead>
<tr>
<th>Configuration*</th>
<th>Shoulder discomfort</th>
<th>Hip discomfort</th>
<th>Overall discomfort</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 (15.0 kg, 30 N)</td>
<td>4.28 ± 0.85</td>
<td>2.34 ± 0.50</td>
<td>3.40 ± 0.53</td>
</tr>
<tr>
<td>2 (15.0 kg, 60 N)</td>
<td>3.47 ± 0.73</td>
<td>2.69 ± 0.48</td>
<td>3.12 ± 0.57</td>
</tr>
<tr>
<td>3 (15.0 kg, 90 N)</td>
<td>2.94 ± 0.58</td>
<td>3.11 ± 0.57</td>
<td>2.94 ± 0.39</td>
</tr>
<tr>
<td>4 (15.0 kg, 120 N)</td>
<td>2.59 ± 0.77</td>
<td>4.61 ± 0.92</td>
<td>3.35 ± 0.66</td>
</tr>
<tr>
<td>5 (20.0 kg, 30 N)</td>
<td>5.15 ± 0.88</td>
<td>2.66 ± 0.65</td>
<td>4.14 ± 0.63</td>
</tr>
<tr>
<td>6 (20.0 kg, 60 N)</td>
<td>4.25 ± 0.78</td>
<td>3.07 ± 0.48</td>
<td>3.90 ± 0.59</td>
</tr>
<tr>
<td>7 (20.0 kg, 90 N)</td>
<td>3.69 ± 0.84</td>
<td>4.03 ± 0.55</td>
<td>3.72 ± 0.42</td>
</tr>
<tr>
<td>8 (20.0 kg, 120 N)</td>
<td>3.75 ± 0.80</td>
<td>5.44 ± 0.78</td>
<td>4.41 ± 0.61</td>
</tr>
<tr>
<td>9 (25.0 kg, 30 N)</td>
<td>6.19 ± 0.70</td>
<td>2.46 ± 0.45</td>
<td>5.13 ± 0.50</td>
</tr>
<tr>
<td>10 (25.0 kg, 60 N)</td>
<td>5.26 ± 0.75</td>
<td>3.99 ± 0.51</td>
<td>4.49 ± 0.52</td>
</tr>
<tr>
<td>11 (25.0 kg, 90 N)</td>
<td>5.26 ± 0.72</td>
<td>4.52 ± 0.58</td>
<td>4.69 ± 0.39</td>
</tr>
<tr>
<td>12 (25.0 kg, 120 N)</td>
<td>4.47 ± 0.84</td>
<td>5.92 ± 0.79</td>
<td>5.77 ± 0.53</td>
</tr>
</tbody>
</table>

* The configurations differ in load mass and tension to which the hip belt was adjusted, as shown in brackets.
**S2 Table. Mechanical parameters in the shoulder region.** For each measured configuration, the mean of all subjects ± the standard error of measurement is shown.

<table>
<thead>
<tr>
<th>Configuration</th>
<th>Shoulder region</th>
<th>Average pressure [kPa]</th>
<th>Peak pressure [kPa]</th>
<th>Strap force [N]</th>
<th>Relative motion [mm/s]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>static</td>
<td>dynamic</td>
<td>static</td>
<td>dynamic</td>
<td>static</td>
</tr>
<tr>
<td>1 (15.0 kg, 30 N)</td>
<td>6.97 ± 0.98</td>
<td>9.28 ± 0.93</td>
<td>12.43 ± 1.73</td>
<td>24.90 ± 2.32</td>
<td>51.93 ± 2.04</td>
</tr>
<tr>
<td>2 (15.0 kg, 60 N)</td>
<td>4.90 ± 1.12</td>
<td>9.89 ± 0.93</td>
<td>9.16 ± 1.87</td>
<td>23.98 ± 1.86</td>
<td>47.96 ± 1.65</td>
</tr>
<tr>
<td>3 (15.0 kg, 90 N)</td>
<td>4.48 ± 1.17</td>
<td>9.41 ± 0.83</td>
<td>7.98 ± 2.21</td>
<td>22.58 ± 1.90</td>
<td>44.99 ± 2.06</td>
</tr>
<tr>
<td>4 (15.0 kg, 120 N)</td>
<td>4.76 ± 1.32</td>
<td>8.60 ± 0.87</td>
<td>7.74 ± 2.20</td>
<td>20.05 ± 1.90</td>
<td>42.29 ± 1.66</td>
</tr>
<tr>
<td>5 (20.0 kg, 30 N)</td>
<td>10.83 ± 1.27</td>
<td>11.69 ± 0.84</td>
<td>24.20 ± 3.22</td>
<td>34.61 ± 2.53</td>
<td>63.23 ± 2.47</td>
</tr>
<tr>
<td>6 (20.0 kg, 60 N)</td>
<td>9.55 ± 1.53</td>
<td>11.50 ± 0.77</td>
<td>20.59 ± 2.35</td>
<td>37.96 ± 4.61</td>
<td>59.26 ± 2.43</td>
</tr>
<tr>
<td>7 (20.0 kg, 90 N)</td>
<td>8.86 ± 1.48</td>
<td>10.98 ± 0.76</td>
<td>18.51 ± 3.13</td>
<td>30.16 ± 2.36</td>
<td>54.13 ± 2.58</td>
</tr>
<tr>
<td>8 (20.0 kg, 120 N)</td>
<td>8.77 ± 1.62</td>
<td>9.82 ± 1.23</td>
<td>18.33 ± 3.56</td>
<td>28.22 ± 3.78</td>
<td>53.21 ± 2.10</td>
</tr>
<tr>
<td>9 (25.0 kg, 30 N)</td>
<td>12.41 ± 0.97</td>
<td>13.15 ± 0.76</td>
<td>33.58 ± 2.71</td>
<td>44.78 ± 4.06</td>
<td>74.21 ± 1.62</td>
</tr>
<tr>
<td>10 (25.0 kg, 60 N)</td>
<td>13.45 ± 0.98</td>
<td>12.73 ± 0.79</td>
<td>30.31 ± 1.89</td>
<td>43.28 ± 3.48</td>
<td>70.48 ± 1.57</td>
</tr>
<tr>
<td>11 (25.0 kg, 90 N)</td>
<td>9.91 ± 1.06</td>
<td>12.65 ± 0.78</td>
<td>25.91 ± 2.72</td>
<td>39.15 ± 2.63</td>
<td>67.77 ± 1.52</td>
</tr>
<tr>
<td>12 (25.0 kg, 120 N)</td>
<td>12.02 ± 1.20</td>
<td>12.32 ± 0.80</td>
<td>25.09 ± 2.11</td>
<td>37.42 ± 2.99</td>
<td>65.69 ± 2.26</td>
</tr>
</tbody>
</table>
S3 Table. Mechanical parameters in the hip region. For each measured configuration, the mean of all subjects ± the standard error of measurement is shown.

<table>
<thead>
<tr>
<th>Configuration</th>
<th>Hip region</th>
<th>Average pressure [kPa]</th>
<th>Peak pressure [kPa]</th>
<th>Strap force [N]</th>
<th>Relative motion [mm/s]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>static</td>
<td>dynamic</td>
<td>static</td>
<td>dynamic</td>
</tr>
<tr>
<td>1 (15.0 kg, 30 N)</td>
<td>1.19 ± 1.78</td>
<td>12.68 ± 1.22</td>
<td>20.92 ± 2.82</td>
<td>31.85 ± 4.42</td>
<td>3.60 ± 0.44</td>
</tr>
<tr>
<td></td>
<td>12.0 ± 1.07</td>
<td>11.40 ± 1.07</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>2 (15.0 kg, 60 N)</td>
<td>14.81 ± 2.31</td>
<td>14.64 ± 1.63</td>
<td>37.74 ± 5.03</td>
<td>58.87 ± 6.41</td>
<td>3.61 ± 0.47</td>
</tr>
<tr>
<td></td>
<td>10.1 ± 1.06</td>
<td>11.49 ± 1.02</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>3 (15.0 kg, 90 N)</td>
<td>16.81 ± 1.94</td>
<td>16.43 ± 2.07</td>
<td>57.53 ± 4.37</td>
<td>77.79 ± 6.63</td>
<td>3.62 ± 0.47</td>
</tr>
<tr>
<td></td>
<td>12.7 ± 1.18</td>
<td>11.14 ± 1.06</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>4 (15.0 kg, 120 N)</td>
<td>18.53 ± 2.07</td>
<td>17.77 ± 2.14</td>
<td>69.84 ± 5.94</td>
<td>97.67 ± 6.06</td>
<td>3.56 ± 0.44</td>
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<tr>
<td></td>
<td>12.8 ± 1.18</td>
<td>11.18 ± 1.04</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>5 (20.0 kg, 30 N)</td>
<td>9.98 ± 1.28</td>
<td>13.04 ± 1.82</td>
<td>22.14 ± 3.57</td>
<td>31.61 ± 3.29</td>
<td>3.61 ± 0.47</td>
</tr>
<tr>
<td></td>
<td>12.7 ± 1.18</td>
<td>11.80 ± 1.33</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>6 (20.0 kg, 60 N)</td>
<td>15.55 ± 1.94</td>
<td>15.79 ± 1.95</td>
<td>41.90 ± 4.86</td>
<td>59.86 ± 6.74</td>
<td>3.53 ± 0.44</td>
</tr>
<tr>
<td></td>
<td>12.7 ± 1.18</td>
<td>11.81 ± 1.19</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>7 (20.0 kg, 90 N)</td>
<td>17.07 ± 1.92</td>
<td>16.80 ± 1.93</td>
<td>57.08 ± 7.11</td>
<td>83.61 ± 6.42</td>
<td>3.58 ± 0.46</td>
</tr>
<tr>
<td></td>
<td>12.7 ± 1.18</td>
<td>11.83 ± 1.24</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>8 (20.0 kg, 120 N)</td>
<td>19.08 ± 2.16</td>
<td>17.30 ± 1.95</td>
<td>78.47 ± 8.45</td>
<td>105.61 ± 5.61</td>
<td>3.56 ± 0.46</td>
</tr>
<tr>
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<td>12.7 ± 1.18</td>
<td>11.76 ± 1.21</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>9 (25.0 kg, 30 N)</td>
<td>12.76 ± 1.97</td>
<td>13.23 ± 1.76</td>
<td>24.73 ± 3.38</td>
<td>38.61 ± 6.30</td>
<td>3.53 ± 0.46</td>
</tr>
<tr>
<td></td>
<td>12.7 ± 1.18</td>
<td>11.91 ± 1.26</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>10 (25.0 kg, 60 N)</td>
<td>16.01 ± 2.26</td>
<td>16.50 ± 2.10</td>
<td>43.16 ± 6.23</td>
<td>64.43 ± 6.02</td>
<td>3.52 ± 0.46</td>
</tr>
<tr>
<td></td>
<td>12.7 ± 1.18</td>
<td>12.06 ± 1.20</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>11 (25.0 kg, 90 N)</td>
<td>17.96 ± 2.24</td>
<td>17.53 ± 2.25</td>
<td>61.67 ± 6.54</td>
<td>86.80 ± 5.46</td>
<td>3.69 ± 0.43</td>
</tr>
<tr>
<td></td>
<td>12.7 ± 1.18</td>
<td>11.99 ± 1.15</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>12 (25.0 kg, 120 N)</td>
<td>19.27 ± 2.33</td>
<td>17.48 ± 2.01</td>
<td>81.54 ± 9.63</td>
<td>108.72 ± 5.21</td>
<td>3.52 ± 0.46</td>
</tr>
<tr>
<td></td>
<td>12.7 ± 1.18</td>
<td>12.12 ± 1.21</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
5. Validation of an Instrumented Dummy to Assess Mechanical Aspects of Discomfort during Load Carriage

Patrick D. Wettenschwiler\textsuperscript{1,2}, Simon Annaheim\textsuperscript{1}, Silvio Lorenzetti\textsuperscript{2}, Stephen J. Ferguson\textsuperscript{2}, Rolf Stämpfli\textsuperscript{1}, Agnes Psikuta\textsuperscript{1}, René M. Rossi\textsuperscript{1}

Affiliation:

\textsuperscript{1} Empa, Swiss Federal Laboratories for Materials Science and Technology, St. Gallen, Switzerland

\textsuperscript{2} Institute for Biomechanics, ETH Zurich, Zurich, Switzerland


Reprinted with approval from all co-authors.
5.1. Abstract

Due to the increasing load in backpacks and other load carriage systems over the last decades, load carriage system designs have to be adapted accordingly to minimize discomfort and to reduce the risk of injury. As subject studies are labor-intensive and include further challenges such as intra-subject and inter-subject variability, we aimed to validate an instrumented dummy as an objective laboratory tool to assess the mechanical aspects of discomfort. The validation of the instrumented dummy was conducted by comparison with a recent subject study. The mechanical parameters that characterize the static and dynamic interaction between backpack and body during different backpack settings were compared. The second aim was to investigate whether high predictive power (coefficient of determination $R^2>0.5$) in assessing the discomfort of load carriage systems could be reached using the instrumented dummy. Measurements were conducted under static conditions, simulating upright standing, and dynamic conditions, simulating level walking. Twelve different configurations of a typical load carriage system, a commercially available backpack with a hip belt, were assessed. The mechanical parameters were measured in the shoulder and the hip region of the dummy and consisted of average pressure, peak pressure, strap force and relative motion between the system and the body. The twelve configurations consisted of three different weights (15kg, 20kg, and 25kg), combined with four different hip belt tensions (30N, 60N, 90N, and 120N). Through the significant ($p<0.05$) correlation of the mechanical parameters measured on the dummy with the corresponding values of the subject study, the dummy was validated for all static measurements and for dynamic measurements in the hip region to accurately simulate the interaction between the human body and the load carriage system. Multiple linear regressions with the mechanical parameters measured on the dummy as independent variables and the corresponding subjective discomfort scores from the subject study as the dependent variable revealed a high predictive power of the instrumented dummy. The dummy can explain 75% or more of the variance in discomfort using average pressures as predictors and even 79% or more of the variance in discomfort using strap forces as predictors. Use of the dummy enables objective, fast, and iterative assessments of load carriage systems and therefore reduces the need for labor-intensive subject studies in order to decrease the mechanical aspects of discomfort during load carriage.
5.2. Introduction

During the last decades, the loads carried in backpacks and other load carriage systems have increased [1, 2]. As a consequence, load carriage system designs have to be adapted accordingly to minimize discomfort and to reduce the risk of injury. Discomfort during load carriage is a major issue relevant for e.g. infantry [3-6], school children’s or student’s load carriage [7-10], and outdoor activities [11-13]. The importance of discomfort is further underlined by its influence on user acceptance [14, 15]. Recently, a growing interest has been shown in the mechanical aspects of discomfort during load carriage: Piscione et al. [16] analyzed the effect of mechanical compression on shoulder muscle fatigue and Hadid et al. [17] modeled the mechanical strains and stresses in the soft tissue of the shoulder. Older publications by Holloway et al. [18] and Sangeorzan et al. [19] already investigated the relationship between external pressure loading and skin blood flow. Furthermore, a number of investigations reported a relationship between discomfort and the external pressure loading occurring during sitting activities, e.g. in car seats [20], or office chairs [21]. While most of these studies investigated static scenarios, the dynamic aspects of the mechanical loading may well play an important role for the perceived discomfort. There are still many unsolved questions regarding the mechanical aspects of discomfort.

Currently, assessing the mechanical aspects of discomfort in load carriage systems is mostly done using subject studies, where the participants provide subjectively perceived discomfort scores. However, these studies are labor-intensive and include further challenges, such as intra-subject and inter-subject variability or ethical considerations for extreme applications. Furthermore, direct comparisons of load carriage systems are often only possible within the same study. Hence, to avoid biased results, an objective measurement tool in a laboratory environment would be preferable, especially when comparing many different load carriage systems. Such a tool would also be beneficial for the industry during different stages of the development of load carriage systems. A first instrumented physical model, developed for this purpose, was presented by Stevenson et al. [5]. For the shoulder region, their model explained 31% of the variation in discomfort using average pressure measured on the model [22]. As a validation, they presented significant correlations (p<0.05) between the mechanical parameters measured on their model and discomfort scores gathered
from subjects [22]. For further development in this field, future models need to enhance the predictive power in assessing discomfort, while at the same time providing a solid model validation. Recently, new possibilities for the validation of such models have opened up. For the first time, mechanical parameters that characterize the static and dynamic interaction between a load carriage system and the human body, have been measured in a subject study [23]. In addition, this study revealed that static peak pressure, or alternatively strap force in the shoulder strap and the hip belt, can account for more than 85% of the variation in subjectively perceived discomfort in the shoulder and the hip region [23]. As a consequence, a physical model that is able to accurately simulate human load carriage, regarding the relevant mechanical parameters, would be a promising tool to assess the mechanical aspects of discomfort of load carriage systems economically, objectively and reliably. Therefore, the first aim of this study was to validate a newly built instrumented dummy by comparing mechanical parameters that characterize the static and dynamic interaction between load carriage system and body, during different backpack settings, with the recent subject study. The second aim of this study was to investigate whether a high predictive power (coefficient of determination $R^2 > 0.5$) in assessing the discomfort of load carriage systems can be reached using the instrumented dummy.

5.3. Materials and Methods

5.3.1. Instrumented Dummy

Based on the model of Stevenson et al. [5], we constructed a dummy to assess mechanical aspects of discomfort. It features a male anatomy with a chest circumference of 104cm, a waist circumference of 81cm, and a waist-to-neck length of 52cm (Fig. 1). As a skin analogue, its body surface consists of a 3.0mm thick layer of nora® Lunasoft SLW (Otto Bock Healthcare, Duderstadt, Germany). This closed cell foam was developed for the interface of human prostheses and has a Shore A hardness of 30, complying with the hardness of human skin and the subcutaneous tissue [24]. To compensate the external loading of the backpack, the natural forward leaning that occurs during posterior load carriage [25-28] is replicated in the dummy. According to Grimmer et al. [25], the flexion during posterior load carriage occurs in the ankle. Therefore, in this study, the forward lean angle in the upper ankle joint is calculated to balance the moment about the upper ankle joint. For this calculation,
the mass distribution of the human body segments of a medium sized male (bodyweight 81.5kg, size 178.4cm) was applied according to Armstrong et al. [29]. The mass (± 0.1kg) and the center of mass (± 0.1cm) of the load carriage system in relation to the upper ankle joint are measured using a trifilied pendulum and a custom made straightedge in the sagittal plane of the dummy. The three different masses of the load carriage system used in this study (see section ‘Load Carriage System’ below) required forward leaning angles of the dummy of 2.10°, 2.64°, and 3.13°, respectively, to achieve equilibrium. For practical reasons, the dummy did not wear clothes. Clothes were shown to have only a minor influence on the pressure distribution on the body surface during load carriage [30]. For the static scenario, the dummy simulates upright standing, while for the dynamic scenario the dummy simulates walking or jogging through a vertical sinusoidal motion with an amplitude of 50.8mm and a frequency of 1.77Hz. This frequency corresponds to a walking speed of approximately 4.5km/h [31]. Theoretically, the dummy can also measure the load distribution between the hip and the upper body during load carriage. However, for this study, the two six degrees of freedom load cells, which were integrated into the dummy for this purpose, were not involved in the measurements. They may serve in future studies, e.g. to analyze the loading of the lumbar spine during load carriage.

Fig. 1. Newly built dummy. Tekscan type 9811E sensors are mounted in the shoulder and the hip region. MFS: magnetic field source of the 3D motion tracking device;
LC1: load cell in the pelvis; LC2: load cell in the upper body; Act: actuator for the sinusoidal motion.

5.3.2. Body Surface Pressure Measurement

The pressure on dedicated regions of the surface of the dummy was recorded using Tekscan type 9811E pressure sensitive foils (Tekscan, South Boston, MA, USA) with a pressure range up to 172kPa (Fig. 1). These sensors feature 6 x 16 sensor cells, covering a total area of 7.6cm x 20.3cm. Thus, a detailed pressure distribution was obtained for each dedicated region. Before the measurements, conditioning, equilibration, and calibration was performed according to the manufacturer’s guidelines. Equilibration was performed at 20kPa, while three-point calibration was performed at 0kPa, 20kPa, and 50kPa. The sensors were placed on the right shoulder and right hip region of the dummy, as shown in Fig. 1. It has been reported that the placement of the pressure sensors on a curved surface may induce artefacts due to bending of the sensors [32]. The signal of the sensors placed on the dummy without the load carriage system mounted was recorded prior to every measurement, but in this study, no artefacts were observed. Therefore, no further offset correction had to be conducted. The data from the pressure sensors was processed in MATLAB (R2012b, The MathWorks, Natick, MA, USA) to determine the average pressure and the peak pressure for each dedicated region. This way, the average and the peak pressure were obtained for all static and dynamic measurements in the shoulder and the hip region. With a measurement duration of 10.2 seconds at a sampling rate of 120Hz, a single measurement consisted of 1224 frames, each including pressure data of 6 x 16 sensor cells. The average pressure value of all cells interacting with the load carriage system was calculated by first taking the mean of all non-zero cells in each frame and subsequently taking the mean of these mean values over time. The peak pressure value was calculated by extracting the overall maximum value over the 6 x 16 cells and over time.

5.3.3. Strap Force Measurement

The forces were measured in the left shoulder strap and in the hip belt using custom made force sensors based on strain gauges (Fig. 2). To calculate the strap forces for each measurement, the mean value over time was calculated.
Fig. 2. Dummy with load carriage system. The strap force sensors are mounted in the shoulder strap and the hip belt (white arrows). The Polhemus LIBERTY source is mounted on top of the dummy and its sensors are at the top and the bottom of the load carriage system (black arrows).

5.3.4. Relative Motion Measurement

Using the 3D motion tracking system Polhemus LIBERTY 240/8 (Polhemus, Colchester, VT, USA) we measured the relative motion between the bulk of the load carriage system and the dummy. The electromagnetic field source of the tracking system was mounted on top of the dummy (Fig. 2). The 6 degrees-of-freedom sensors were mounted on the left side of the load carriage system, one at the top and one at the bottom (Fig. 2). For the relative motion in each measurement, the cumulative change of distance between the sensor and the source in all three axes was calculated. To normalize the relative motion over time, the cumulative change of distance was divided by the measurement duration.
5.3.5. Load Carriage System

The commercially available backpack “Deuter ACT Lite 50+10” (Deuter Sport GmbH, Gersthofen, Germany) served as load carriage system in this study. Its intended use ranges from travelling to trekking and alpine tours. Due to the use of the electromagnetic tracking system, no metal parts were allowed in the load carriage system. Even the paramagnetic aluminum rods that formed the frame in the back wall of the backpack had to be removed. Instead, a wooden box (65.0cm height, 27.5cm length, 15.8cm depth, and 0.9cm wall thickness) was inserted into the backpack and rigidly fixed to the back wall, replacing the function of the aluminum rods and thus enabling a load transfer between the hip belt and the shoulder straps. Rather than applying several different load carriage systems, we applied twelve different configurations of this one typical load carriage system, thus eliminating potential effects of the design on discomfort. As load weight and hip belt tension have the largest effects on interface pressure and strap forces [33], the twelve configurations used in our study consisted of a combination of three different loads and four different hip belt tensions: The hip belt lengths were adjusted to reach 30N, 60N, 90N, and 120N (±1N) of static tension on the dummy. The total masses of the load carriage system configurations were 15.0kg, 20.0kg, and 25.0kg. For an efficient change of the payload, cardboard boxes filled with sand could be inserted through an opening in the backpack and the wooden box at the bottom and the top (Fig. 3). All configurations respected lateral symmetry, while the center of mass was positioned 30.5cm from the bottom in all configurations.
Fig. 3. **Load carriage system.** Cardboard boxes filled with sand could be inserted through an opening in the backpack and the wooden box at the bottom and the top to enable efficient change of the payload, while keeping the center of mass in the same position.

### 5.3.6. Experimental Protocol

The static and the dynamic simulations on the instrumented dummy were repeated five times and the mean values are presented for all mechanical parameters. The measurement duration was 10.2 seconds for the static and the dynamic condition, resulting in 18 simulated steps (at 1.77Hz) in the dynamic measurements. For all mechanical parameters, the sampling rate was 120Hz. For each simulation, the load carriage system was removed and remounted on the dummy. We then compared the results of the dummy measurements with the results of the previous subject study [23], in which the same mechanical parameters were assessed in the same way, along with the subjectively perceived discomfort of the subjects. The previous subject study applied the same load carriage system with the same twelve configurations.
5.3.7. Statistical Analysis

For a direct validation of the dummy, we compared the mechanical parameters measured on the dummy with the same parameters previously measured in the subject study [23]. Spearman’s rho was calculated for each mechanical parameter, indicating how well the relationship between two variables can be described using a monotonic function. A nonparametric correlation coefficient had to be chosen, as the dynamic average pressure in the hip, measured on the subjects, was not normally distributed according to the Shapiro-Wilk test ($p=0.047$).

We further investigated how much variation in subjectively perceived discomfort could be explained by our instrumented dummy. For the shoulder and the hip region each, a first round of multiple linear regressions was conducted with the subjective discomfort from the previous subject study [23] as the dependent variable. The independent variables were the pressure parameters measured on the dummy, i.e. static average pressure, static peak pressure, dynamic average pressure, and dynamic peak pressure. The pressure parameters provide information independent of the type of a load carriage system, because they can be measured directly on the body surface of the dummy. In order to investigate whether more variance in discomfort could be explained by additionally including the strap forces and the relative motion as independent variables, a second round of multiple linear regressions was performed. Due to the dependence on the type and design of a load carriage system, results including strap forces and relative motion as predictors may only be applied to load carriage systems of the specific type of backpack with a hip belt. In the second round of regressions, instead of conducting multiple linear regressions with eleven degrees of freedom and eight possible predictors, separate regressions were conducted for the static and dynamic scenarios. All regression analyses were conducted using backwards elimination method. Significance was defined as $p<0.05$ and Bonferroni correction for multiple testing was applied on all analyses.

5.4. Results

The correlation coefficients comparing the mechanical parameters between the dummy and the subjects are shown in Table 1.
The result of the multiple linear regression analyses with discomfort as dependent variable and the pressure parameters as independent variables is shown in Fig. 4 for the shoulder region and in Fig. 5 for the hip region.

The results of the multiple linear regression analyses with all mechanical parameters as independent variables are shown in Table 2. When strap force was included, the pressure parameters could not explain any significant additional variance and were excluded from the models due to the backwards elimination method (Table 2).

**Table 1: Correlation coefficients between the mechanical parameters measured on the subjects and on the dummy.** *+ non-normal distribution (p=0.047) of the subjects' variable according to Shapiro-Wilk test. * p<0.05, ** p<0.01, *** p<0.001*

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Spearman-Rho</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulder static average pressure</td>
<td>0.87***</td>
</tr>
<tr>
<td>Shoulder static peak pressure</td>
<td>0.62*</td>
</tr>
<tr>
<td>Shoulder static strap force</td>
<td>0.86***</td>
</tr>
<tr>
<td>Shoulder dynamic average pressure</td>
<td>not significant</td>
</tr>
<tr>
<td>Shoulder dynamic peak pressure</td>
<td>not significant</td>
</tr>
<tr>
<td>Shoulder dynamic strap force</td>
<td>0.86***</td>
</tr>
<tr>
<td>Shoulder dynamic relative motion</td>
<td>not significant</td>
</tr>
<tr>
<td>Hip static average pressure</td>
<td>0.94***</td>
</tr>
<tr>
<td>Hip static peak pressure</td>
<td>0.86***</td>
</tr>
<tr>
<td>Hip static strap force</td>
<td>0.92***</td>
</tr>
<tr>
<td>Hip dynamic average pressure†</td>
<td>0.93***</td>
</tr>
<tr>
<td>Hip dynamic peak pressure</td>
<td>0.81**</td>
</tr>
<tr>
<td>Hip dynamic strap force</td>
<td>0.99***</td>
</tr>
<tr>
<td>Hip dynamic relative motion</td>
<td>0.87***</td>
</tr>
</tbody>
</table>
Fig. 4. Results of the regression analysis using the pressure parameters measured on the dummy as independent variables. Static average pressure measured on the dummy is a significant (p<0.001) predictor of the subjects’ discomfort in the shoulder region. The subject’s discomfort was rated on a visual analogue scale, ranging from 0 = no discomfort to 10 = maximal discomfort. The non-significant predictors were removed from the model during the backwards elimination of the regression analysis. Data points show the mean ± standard error of measurement for each configuration, dotted lines show 95% prediction intervals. Regression equation: y = 1.512x - 1.044, $R^2 = 0.82$.

Fig. 5. Results of the regression analysis using the pressure parameters measured on the dummy as independent variables. Dynamic average pressure measured on the dummy as independent variables.
sured on the dummy is a significant (p<0.001) predictor of the subjects’ discomfort in the hip region. Data points show the mean ± standard error of measurement for each configuration, dotted lines show 95% prediction intervals. Regression equation: y = 0.999x + 0.041, R² = 0.75.

Table 2: Results of the regression analyses using average pressure, peak pressure, strap force, and relative motion measured on the dummy as independent variables. The dependent variable was the subjects’ discomfort. Each resulting model consists of an equation to predict the subjectively perceived discomfort through a constant plus the model parameters multiplied by their coefficients. Relative motion was only entered as independent variable for the dynamic conditions.

*p<0.01, ** p<0.001

<table>
<thead>
<tr>
<th>Condition</th>
<th>Region</th>
<th>R²</th>
<th>Model</th>
<th>Coefficients</th>
<th>95% confidence interval of coefficients</th>
<th>Standardized coeff.</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>lower</td>
<td>upper</td>
</tr>
<tr>
<td>Static</td>
<td>Shoulder</td>
<td>0.88</td>
<td>Constant</td>
<td>0.855</td>
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<td>1.777</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Strap force</td>
<td>0.116</td>
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<td>0.146</td>
</tr>
<tr>
<td></td>
<td>Hip</td>
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<td>Constant</td>
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<td>0.505</td>
<td>2.346</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Strap force</td>
<td>0.031</td>
<td>0.020</td>
<td>0.042</td>
</tr>
<tr>
<td>Dynamic</td>
<td>Shoulder</td>
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<td>Constant</td>
<td>1.032</td>
<td>0.252</td>
<td>1.812</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Strap force</td>
<td>0.106</td>
<td>0.081</td>
<td>0.130</td>
</tr>
<tr>
<td></td>
<td>Hip</td>
<td>0.94</td>
<td>Constant</td>
<td>-5.429</td>
<td>-8.925</td>
<td>-1.933</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Strap force</td>
<td>0.032</td>
<td>0.025</td>
<td>0.038</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Relative motion</td>
<td>4.420</td>
<td>2.177</td>
<td>6.663</td>
</tr>
</tbody>
</table>

5.5. Discussion

In this study, we aimed to validate an instrumented dummy to assess the mechanical aspects of discomfort during load carriage by comparing the mechanical parameters that characterize the interaction between the body and the load carriage system with corresponding values from a previous subject study. We further analyzed the predic-
tive power of the instrumented dummy in assessing the discomfort of load carriage systems, applying twelve different backpack configurations.

The mechanical parameters measured on the dummy show significant (p<0.05) correlations (Spearman’s rho>0.62) with the corresponding parameters measured on the subjects, except for the dynamic measurements in the shoulder region (Table 1). This proves the validity of the dummy for all static measurements and for dynamic measurements in the hip region to accurately simulate the interaction between the human body and the load carriage system. In the shoulder region, the pressure parameters and the relative motion revealed low agreement for the dynamic measurements. This may be due to the rigid connection of the pelvis and the shoulder girdle in the dummy, whereas the motion pattern of the subjects naturally includes a counter-rotation of the pelvis and the shoulder girdle [34-36]. Nevertheless, the shoulder strap force did correlate highly (Spearman’s rho =0.86) in the shoulder region (Table 1). Therefore, the dynamic scenario in the shoulder region can still be assessed on the dummy with any load carriage system that feature shoulder straps and has a similar design as the system used in this study.

Regarding the predictive power of the instrumented dummy, positive results were found already for the multiple linear regression analyses with only the pressure parameters as independent variables. In the shoulder region, the static average pressure measured on the dummy can explain 82% of the variance in the subjectively perceived discomfort (Fig. 4). In the hip region, the dynamic average pressure measured on the dummy can explain 75% of the variance in the subjectively perceived discomfort (Fig. 5). Regarding the measurements on the dummy, average pressure is therefore a better predictor of discomfort than peak pressure, but it is not clear whether static or dynamic average pressure is more appropriate. In the shoulder region, dynamic average pressure on the dummy and on the subjects was not significantly correlated (Table 1), which makes dynamic average pressure a less suitable predictor of discomfort. In the hip region however, the correlations between the measurements on the dummy and on the subjects are similarly high for static and dynamic average pressure (Table 1). It is possible that in the absence of dynamic average pressure, static average pressure could be a significant predictor of discomfort in the hip region. Among all the mechanical parameters measured on the instrumented dummy, the pressure parameters are the ones that provide information independent
of the load carriage system type, because they are measured directly on the body surface of the dummy. Therefore, the average pressure can be regarded as a significant (p<0.001) and a very powerful predictor of discomfort, regardless of the type of load carriage system evaluated.

When considering all mechanical parameters measured on the dummy, multiple linear regression analyses revealed the strap force to be a significant (p<0.001) predictor of discomfort, accounting for 79% or more of the variance in the subjectively perceived discomfort (Table 2). Strap force was more significant as a predictor than average pressure and average pressure could not explain any significant additional variance. Therefore, average pressure was excluded from the models during the backwards elimination (Table 2). For the dynamic scenario, the relative motion was also a significant (p<0.01) predictor of discomfort in the hip region. But the lower significance level of relative motion, along with its smaller standardized coefficient and the large confidence interval of its coefficient in the regression model (Table 2), indicate that strap force is more relevant. Both strap force and relative motion may depend on the type and design of a load carriage system. As all measurements in this study have been conducted with a load carriage system of a backpack with a hip belt, the relevance of strap force and relative motion as predictors of discomfort must currently be restricted to load carriage systems of comparable type and design. Most importantly, all mechanical parameters measured on the dummy that were identified as significant predictors of discomfort, are significantly (p<0.001) correlated to the corresponding parameters measured on subjects (Table 1).

Compared to the existing model of Stevenson et al. [5, 22], our instrumented dummy exhibits a higher predictive power and uses mechanical parameters as predictors that are directly validated by correlation with the same parameters from a human subject study. This direct validation is one of the crucial features of the instrumented dummy presented in this study. Its uniqueness is further based on its pure focus on the mechanical aspects of discomfort. While thermo-physiological aspects play an important role in discomfort during load carriage, already existing models from the field of clothing physiology offer convenient possibilities to test load carriage systems under various climatic conditions [37, 38].
The average pressure values measured on the dummy (Fig. 4 and Fig. 5) are mostly lower than the values shown to be critical for skin blood flow occlusion (5.6-16kPa), as reported by Holloway et al. [18] or Sangeorzan et al. [19]. These findings are in line with the discomfort scores not reaching the highest levels, as reported by the subjects wearing the same load carriage system configurations [23]. However, the pressure values measured on the dummy are lower than the corresponding values measured on the subjects, e.g. the static average pressure in the shoulder region ranged from 2.7kPa to 4.7kPa in the dummy measurements and from 4.5kPa to 13.5kPa in the subject study [23]. The absolute values measured on the dummy must therefore be treated with care. Because the Tekscan sensors can bend on the body surface of the subjects during dynamic measurements, the absolute values measured on the subjects may be overestimated by artefacts, despite minimizing this effect through offset correction. By contrast, the dummy has a rigid body, which holds less potential for dynamic bending of the Tekscan sensors. This may partly explain the difference between the absolute pressure values measured on the subjects and on the dummy. While the deviation from reality in terms of arm motion or torsion of the dummy is a limitation, it enables more reliable and more objective measurements than subject studies. Compared to subject studies, the dummy is also more objective in case of potentially confounding factors. Effects of the aesthetic design of a load carriage system or personal moods are excluded. Due to the reproducible test conditions, future measurements of load carriage systems can also be directly compared with existing data, resulting from previous measurements on the instrumented dummy.

Generally, the validation of the instrumented dummy presented in this study reduces the need for subject studies to assess discomfort during load carriage. The dummy enables objective, fast, and iterative assessments during the development of new designs of load carriage systems as well as comparisons of many different systems or designs. As a consequence, the users of backpacks and other load carriage systems may profit from future load carriage systems that inflict less discomfort during load carriage.
5.5.1. Limitations

The body surface regions investigated in this study are currently limited to the shoulder and hip region, as they are considered to be the most critical regarding discomfort during load carriage. To allow the application of our method to all relevant body regions, further measurements may be necessary. Another limitation of this study is the male anatomy of the dummy. Gender specific differences in the mechanical parameters and in the perception of discomfort during load carriage are likely and express a need for future work focusing on female subjects and dummies with female anatomy. Furthermore, the forward leaning angles of the dummy are calculated based on the assumption of balancing the moment in the upper ankle joint and possible adaptations in other joints are neglected.

5.6. Conclusions

In this study, an objective and time-saving method to assess or compare the discomfort of load carriage systems and their design has been validated. The instrumented dummy can explain 75% or more of the variance in discomfort using average pressures as predictors and even 79% or more of the variance in discomfort using strap forces as predictors. Compared to an existing model [5], our model has a higher predictive power and it uses mechanical parameters as predictors that are directly validated by correlation with the same parameters measured in a human subject study. Our model enables objective, fast, and iterative assessments during the development of new designs of load carriage systems as well as comparisons of many different systems or designs. As a consequence, the need for subject studies is reduced and users of backpacks and other load carriage systems may profit from future load carriage systems that inflict less discomfort during load carriage.

5.7. Acknowledgments

We thank armasuisse for their financial support. We would also like to thank Dr. Joan Stevenson and her research group from the Biomechanics and Ergonomics Lab at Queen’s University, Canada, for their consultation on building the instrumented dummy.
5.8. References


## 5.9. Supporting Information

**S1 Table. Mechanical parameters in the shoulder region of the instrumented dummy.** For each measured configuration, the mean of five iterations ± the standard error of measurement is shown.

<table>
<thead>
<tr>
<th>Shoulder region</th>
<th>Average pressure [kPa]</th>
<th>Peak pressure [kPa]</th>
<th>Strap force [N]</th>
<th>Relative motion [mm/s]</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Configuration</strong></td>
<td><strong>static</strong></td>
<td><strong>dynamic</strong></td>
<td><strong>static</strong></td>
<td><strong>dynamic</strong></td>
</tr>
<tr>
<td>1 (15.0 kg, 30 N)</td>
<td>3.5 ± 0.1</td>
<td>3.6 ± 0.1</td>
<td>22.5 ± 0.7</td>
<td>23.9 ± 0.6</td>
</tr>
<tr>
<td>2 (15.0 kg, 60 N)</td>
<td>3.0 ± 0.1</td>
<td>3.1 ± 0.1</td>
<td>15.7 ± 0.7</td>
<td>17.0 ± 0.6</td>
</tr>
<tr>
<td>3 (15.0 kg, 90 N)</td>
<td>2.7 ± 0.1</td>
<td>2.7 ± 0.1</td>
<td>11.3 ± 0.7</td>
<td>12.1 ± 0.6</td>
</tr>
<tr>
<td>4 (15.0 kg, 120 N)</td>
<td>3.0 ± 0.1</td>
<td>2.8 ± 0.1</td>
<td>14.6 ± 0.7</td>
<td>14.7 ± 0.6</td>
</tr>
<tr>
<td>5 (20.0 kg, 30 N)</td>
<td>3.8 ± 0.1</td>
<td>3.9 ± 0.1</td>
<td>20.0 ± 1.2</td>
<td>22.0 ± 1.3</td>
</tr>
<tr>
<td>6 (20.0 kg, 60 N)</td>
<td>3.6 ± 0.1</td>
<td>3.5 ± 0.2</td>
<td>20.3 ± 1.0</td>
<td>21.0 ± 0.9</td>
</tr>
<tr>
<td>7 (20.0 kg, 90 N)</td>
<td>3.2 ± 0.1</td>
<td>2.9 ± 0.1</td>
<td>16.2 ± 1.3</td>
<td>15.6 ± 1.0</td>
</tr>
<tr>
<td>8 (20.0 kg, 120 N)</td>
<td>3.0 ± 0.1</td>
<td>2.9 ± 0.1</td>
<td>15.7 ± 1.2</td>
<td>16.1 ± 1.2</td>
</tr>
<tr>
<td>9 (25.0 kg, 30 N)</td>
<td>4.7 ± 0.2</td>
<td>4.2 ± 0.3</td>
<td>23.8 ± 2.0</td>
<td>21.3 ± 1.5</td>
</tr>
<tr>
<td>10 (25.0 kg, 60 N)</td>
<td>4.6 ± 0.2</td>
<td>4.6 ± 0.1</td>
<td>23.2 ± 0.5</td>
<td>23.2 ± 0.7</td>
</tr>
<tr>
<td>11 (25.0 kg, 90 N)</td>
<td>3.9 ± 0.1</td>
<td>3.6 ± 0.1</td>
<td>19.0 ± 0.5</td>
<td>19.1 ± 0.6</td>
</tr>
<tr>
<td>12 (25.0 kg, 120 N)</td>
<td>3.2 ± 0.2</td>
<td>2.4 ± 0.1</td>
<td>13.4 ± 0.7</td>
<td>14.2 ± 0.3</td>
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</tbody>
</table>
**S2 Table. Mechanical parameters in the hip region of the instrumented dummy.** For each measured configuration, the mean of five iterations ± the standard error of measurement is shown.

<table>
<thead>
<tr>
<th>Configuration</th>
<th>Hip region</th>
<th>Average pressure [kPa]</th>
<th>Peak pressure [kPa]</th>
<th>Strap force [N]</th>
<th>Relative motion [mm/s]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>static</td>
<td>dynamic</td>
<td>static</td>
<td>dynamic</td>
</tr>
<tr>
<td>1 (15.0 kg, 30 N)</td>
<td>static</td>
<td>2.2 ± 0.1</td>
<td>2.1 ± 0.1</td>
<td>9.0 ± 0.3</td>
<td>9.0 ± 0.3</td>
</tr>
<tr>
<td>2 (15.0 kg, 60 N)</td>
<td>static</td>
<td>3.4 ± 0.1</td>
<td>3.4 ± 0.1</td>
<td>13.5 ± 0.2</td>
<td>13.6 ± 0.2</td>
</tr>
<tr>
<td>3 (15.0 kg, 90 N)</td>
<td>static</td>
<td>3.9 ± 0.2</td>
<td>3.9 ± 0.2</td>
<td>18.6 ± 1.3</td>
<td>18.8 ± 1.4</td>
</tr>
<tr>
<td>4 (15.0 kg, 120 N)</td>
<td>static</td>
<td>5.5 ± 0.5</td>
<td>5.2 ± 0.4</td>
<td>27.2 ± 2.5</td>
<td>27.6 ± 2.5</td>
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<tr>
<td>5 (20.0 kg, 30 N)</td>
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<td>2.3 ± 0.1</td>
<td>2.5 ± 0.1</td>
<td>10.4 ± 0.6</td>
<td>10.9 ± 0.6</td>
</tr>
<tr>
<td>6 (20.0 kg, 60 N)</td>
<td>static</td>
<td>3.6 ± 0.2</td>
<td>3.6 ± 0.2</td>
<td>20.8 ± 2.1</td>
<td>20.9 ± 2.1</td>
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<tr>
<td>7 (20.0 kg, 90 N)</td>
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<td>3.8 ± 0.3</td>
<td>3.9 ± 0.3</td>
<td>25.2 ± 1.8</td>
<td>25.8 ± 1.9</td>
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<tr>
<td>8 (20.0 kg, 120 N)</td>
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<td>5.2 ± 0.4</td>
<td>5.2 ± 0.4</td>
<td>27.7 ± 2.2</td>
<td>28.1 ± 2.1</td>
</tr>
<tr>
<td>9 (25.0 kg, 30 N)</td>
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<td>2.4 ± 0.1</td>
<td>2.5 ± 0.1</td>
<td>11.5 ± 0.6</td>
<td>12.6 ± 0.6</td>
</tr>
<tr>
<td>10 (25.0 kg, 60 N)</td>
<td>static</td>
<td>3.0 ± 0.2</td>
<td>3.1 ± 0.2</td>
<td>13.5 ± 1.1</td>
<td>14.2 ± 1.1</td>
</tr>
<tr>
<td>11 (25.0 kg, 90 N)</td>
<td>static</td>
<td>4.0 ± 0.1</td>
<td>4.1 ± 0.1</td>
<td>16.9 ± 0.4</td>
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</tr>
<tr>
<td>12 (25.0 kg, 120 N)</td>
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<td>4.9 ± 0.1</td>
<td>21.7 ± 1.0</td>
<td>22.0 ± 0.7</td>
</tr>
</tbody>
</table>
6. Loading of the Lumbar Spine during Backpack Carriage

Patrick D. Wettenschwiler\textsuperscript{1,2}, Silvio Lorenzetti\textsuperscript{2}, Stephen J. Ferguson\textsuperscript{2}, Rolf Stämpfli\textsuperscript{1}, Ameet K. Aiyangar\textsuperscript{3}, René M. Rossi\textsuperscript{1}, Simon Annaheim\textsuperscript{1}

\textsuperscript{1} Empa, Swiss Federal Laboratories for Materials Science and Technology, Lерchenfeldstrasse 5, CH-9014 St. Gallen, Switzerland

\textsuperscript{2} Institute for Biomechanics, ETH Zurich, Hönggerbergring 64, CH-8093 Zürich, Switzerland

\textsuperscript{3} Empa, Swiss Federal Laboratories for Materials Science and Technology, Überlandstrasse 129, CH-8600 Dübendorf, Switzerland


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6.1. Abstract

The use of backpacks and other load carriage systems is significantly associated with a higher prevalence of low back pain. Elevated compression and shear forces in the lumbar intervertebral discs are known risk factors. To avoid invasive measurements, numerical models are often used to predict the relevant forces, but such models face two major challenges: the forces and moments resulting from the interaction between the backpack and body have to be known and the lumbar curvature has to be defined. In this study, a novel method of calculating the loads in the lumbar spine during backpack carriage is presented by combining physical and numerical modeling. The aim of this work was to predict lumbar spine loading during backpack carriage and to investigate the influence of precise lumbar curvature data in numerical modeling of the lumbar compression and shear forces for the conditions of no load, a 20kg and a 40kg backpack. The physical model provided the numerical model with the external forces from the backpack acting on the upper body. To predict realistic lumbar compression forces, subject-specific lumbar curvature data were not necessary for loads up to 40kg. Instead, the fixed lumbar curvature of the generic rigid body model was kept and for the 20kg and 40kg load condition combined with a typical forward lean. In contrast, regarding shear forces, using subject-specific lumbar curvature data from upright MRI measurements as input for the rigid body model significantly altered lumbar joint force estimates. Therefore, whenever the total loading of the spine is of interest, subject-specific lumbar curvature data should be incorporated into spinal models.
6.2. Introduction

The use of backpacks and other load carriage systems is significantly associated with a higher prevalence of low back pain in a wide range of applications, like school children’s load carriage [1] or infantry [2]. The cumulative lumbar compression force and the peak shear force at the L4-L5 intervertebral disc are known risk factors [3]. Relying on subjective perception like discomfort or pain during load carriage [4] as warning signs is not sufficient, as damage of lumbar spine structures may already occur before pain is perceived [5]. Numerical simulations are often used to predict the relevant forces, as non-invasive measurements of these forces in human subjects are impossible. However, there are two major challenges to the development of accurate models: the forces and moments resulting from the mechanical interaction between the backpack and body have to be known and the lumbar curvature has to be defined. To define the lumbar curvature in the models, some researchers apply their own algorithm, minimizing the balancing moments in the lumbar spine [6, 7]. More common is the use of skin markers [8, 9], despite the fact that relative motion between the skin and the underlying bones can introduce errors [10]. An interesting alternative to capture lumbar curvature data has been presented in the last years in the form of upright MRI measurements [11, 12]. This method is limited to static scenarios, but it allows more accurate acquisition of the lumbar curvature than skin markers. The possibility of upright standing during the data acquisition enables the measurement of load carriage scenarios with the correct direction of gravity. However, it is currently unclear how relevant precise lumbar curvature data are for the modeling of the lumbar compression and shear forces during load carriage. This could of course depend on the load carried. A possible step towards a better understanding is to compare the outcome of numerical simulations that use subject-specific lumbar curvature data from upright MRI measurements as input with such that use a fixed lumbar curvature. Therefore, the aim of this study was to predict loading of the lumbar spine during backpack carriage and to further investigate the influence of precise lumbar curvature data in numerical modeling of the lumbar compression and shear forces for the conditions of no load, a 20kg and a 40kg backpack. In order to overcome the challenge of modeling the interaction between backpack and body correctly, we propose to measure the mechanical interaction on a physical model and then apply the resulting forces and moments on a numerical model that was specifically designed to predict lumbar disc forces [13].
6.3. Materials and Methods

6.3.1. Participants

The subject population consisted of nine males without prior occurrence of low back pain and with the following anthropometrics (means ± standard deviation): age 27.7 ± 3.4 years, bodyweight 76.0kg ± 8.4kg, height 177.8cm ± 5.6cm. All subjects provided written, informed consent. This study was approved by the Ethics Commission of the ETH Zurich under the number EK 2015-N-07 and was carried out in accordance with “The Code of Ethics of the World Medical Association” (Declaration of Helsinki, amended October 2013).

6.3.2. Rigid Body Modeling

To predict the intervertebral reaction forces in three different load conditions (0kg = no load, 20kg, 40kg backpack), we applied an OpenSim model that was assembled and enhanced by Senteler et al. [13] based on previous OpenSim models [14-17]. It represents an entire upper human body model, featuring femora, pelvis, sacrum, five individual lumbar vertebrae, a rigid torso (T1-T12 and ribcage), and 83 other rigid bodies comprising neck, head, and arms. In addition to the 238 muscle fascicles introduced by Christophy et al. [14], Senteler et al. [13] incorporated motion segment stiffness in the lumbar spine (L1-S1). At each level, the stiffness contribution of the intervertebral discs, ligaments and capsules is modeled by a linear six degrees of freedom bushing element. Through comparison with in vivo literature data [18-20], Senteler et al. [13] validated the model for the prediction of the lumbar intervertebral joint forces during static tasks with the limitation that postures involving lumbar extension may lead to an overestimation of the joint reaction forces, possibly due to the missing facet joints [13].

For the simulations with adapted lumbar curvature, a size- and mass-scaled subject-specific model was built in OpenSim for each of the subjects. The size-scaling was conducted based on marker positions, using markers at the corner of each vertebral body (T12-L5 and top of S1) in the mid-sagittal plane (Fig. 1a). The two-dimensional subject-specific data for the size-scaling was obtained from the upright MRI results in the 0kg condition. The mass-scaling was based on the total bodyweight of the subjects, while keeping the mass distribution of the template model constant. Using the
scaled models, the OpenSim inverse kinematics tool was applied for each condition (0kg, 20kg, 40kg) using the corresponding marker coordinates from the upright MRI outputs, adjusting the position of the rigid bodies in the sagittal plane. Finally, the standard OpenSim optimization algorithm was used to calculate the muscle recruitment, minimizing the sum of the squared muscle activations. The intervertebral compression and shear forces were expressed in the joint center’s reference frame, whose orientation corresponded to the average of the adjacent vertebral body reference frames [13].

For the simulations with fixed lumbar curvature, a typical forward lean angle was defined for the 20kg and 40kg conditions [21]. The angles were calculated by balancing the moment around the upper ankle joint resulting from the center of mass of the backpack and that of the body. All angles except the upper ankle joint were kept constant. The resulting forward lean angles (2.7° with 20kg, 4.3° with 40kg) are in line with previously published values [22, 23]. For the defined posture, muscle activation and the resulting compression and shear forces were calculated as described above.

The effect of the backpack was modeled by applying external forces on the rigid torso. The size and direction of the external forces corresponded to the forces acting on the torso as measured on an instrumented dummy (see “Physical Modeling”). The point of application of the forces was calculated from the measured moment and the relative position between the load cell in the physical model and the torso-L1 joint in the numerical model.
6.3.3. Upright MRI

All subject measurements were conducted in the institute of radiology in Zürich-Altstetten, Switzerland, using a Fonar Upright™ MRI (Fonar Corporation, Melville, NY, USA). T2-weighted images of the lumbar spine in the mid-sagittal plane were taken during upright standing, first without load, then with the 20kg and 40kg backpack in randomized order. The coordinates of the vertebral body corners (Fig. 1b) were extracted using ImageJ 1.49v (ImageJ, U. S. National Institutes of Health, Bethesda, MD, USA). Due to unintentional motion during the measurement, the images of two subjects were of inadequate quality, making a localization of the vertebral body corners impossible. The exclusion of these images from the study reduced the subject sample to seven subjects.

6.3.4. Physical Modeling

The dummy (physical model) was specifically designed to evaluate load carriage systems [24]. The dummy is horizontally divided at the level of the umbilicus, which corresponds approximately to the level of L4 (Fig. 2). Two AMTI MC2.5B-2K load cells in combination with AMTI Gen 5 signal amplifiers (Advanced Mechanical Technology Inc., Watertown, MA, USA) measure forces in all six degrees of freedom on the upper...
body and the pelvis separately. Due to the focus on the lumbar spine, only the load on the body above the pelvis was of interest. The backpack did not exhibit any interaction with the dummy in the region of the lumbar spine, so the forces and moments measured by the upper load cell were acting on the torso above the lumbar spine. The forward lean angles described above were also applied on the dummy (2.7° with 20kg, 4.3° with 40kg). When the correct angle was set, the load cells were set to zero. Static measurements were conducted five times, taking the backpack off in between and readjusting the position of the shoulder straps and the hip belt. Each measurement lasted 10 seconds with a sampling frequency of 120Hz. First the average of each measurement was calculated, then the average and standard deviations of the five iterations.

![Image of instrumented dummy with backpack](image)

**Fig. 2. Instrumented dummy with backpack.** Applying a typical forward lean angle, the forces and moments acting on the upper body were measured statically. LC: load cell.

### 6.3.5. Backpack

A modified version of the commercially available backpack “Deuter ACT Lite 50+10” (Deuter Sport GmbH, Gersthofen, Germany) served as backpack in this study (Fig. 2). The 0kg, 20kg, and 40kg conditions corresponded to the total mass of the back-
pack, so 0kg meant no backpack. The following modifications were necessary to exclude any metal parts and to create room for the non-metallic payload: The metal rods in the back wall of the backpack were removed. To replace this rigid frame, a wooden box (65.0cm height, 27.5cm length, 15.8cm depth, and 0.9cm wall thickness) was inserted into the backpack and rigidly fixed to the back wall. Another wooden box (16.5cm height, 27.5cm length, 19.0cm depth, and 0.9cm wall thickness) was attached to the bottom to reach the desired payload, which was provided by cardboard boxes filled with glass powder. The center of mass respected lateral symmetry and was located 46.0cm from the lower end and 8.5cm from the back wall for both load conditions. Prior to the measurements, the length of the hip belt was adjusted on the subject or dummy to reach 30N ± 1N of tension, using calibrated customized force sensors based on strain gauges (HBM Typ 1.5/350 XY31, Hottinger Baldwin Messtechnik GmbH, Darmstadt, Germany) and amplifiers (Endevco DC Amplifier Model 136, Meggit Sensing Systems, Irvine, California, USA).

6.3.6. Statistical Analysis

Two-factorial repeated measures ANOVAs were conducted for compression and shear forces separately, to investigate the effects of backpack load (0kg, 20kg, 40kg), lumbar curvature adjustment (adapted, fixed) and their interaction. Where the effect of the lumbar curvature adjustment or the interaction was significant, planned Helmert contrasts compared the 0kg condition with the other two and the 20kg with the 40kg condition. Significance was defined at p<0.05.

6.4. Results

The external loading of the upper body by the backpack was measured on the physical model and is provided in Fig. 3. These forces simulated the external loading in the numerical model. The simulations with adapted and fixed lumbar curvature predicted the compression and shear forces at the L4-L5 level as shown in Fig. 4. The corresponding values for all lumbar levels are displayed in Table 1.

The effect of the backpack load was significant (p<0.001) for the compression and shear forces at all lumbar intervertebral disc levels, except for the shear forces at the L2-L3 level. Where this effect was significant, Helmert contrasts revealed significant
(p<0.001) differences between the 0kg condition and the other two, as well as between the 20kg and the 40kg condition. The effect of the lumbar curvature adjustment was significant (p<0.05) for the shear forces at the L4-L5, L3-L4, and L2-L3 level. The interaction effect of load and lumbar curvature adjustment was significant (p<0.05) for the shear forces at L3-L4, L2-L3, and L1-L2 level. At these levels, Helmert contrasts revealed significant (p<0.05) differences between the 0kg condition and the other two.

Fig. 3. Compression and shear forces and sagittal moment measured between the upper body and the pelvis of the physical model. The error bars indicate standard deviations. Positive shear forces correspond to anterior forces on the upper body. A positive moment corresponds to a lumbar flexion moment.
Fig. 4. Compression (a) and shear force (b) at the L4-L5 level. Light grey: results from simulations with adapted lumbar curvature from upright MRI data. Dark grey: results from simulations using fixed lumbar curvature with typical trunk forward lean. The error bars indicate 95% confidence intervals. Positive shear forces correspond to anterior forces at the cranial end of the intervertebral disc.

* Non-overlapping 95% confidence intervals. ** Significant (p<0.001) Helmert contrasts.
Table 1. Compression and shear forces at all lumbar levels. Mean and 95% confidence intervals (CI) for subject-specific simulations with adapted lumbar curvature (ALC) and fixed lumbar curvature (FLC) simulations, as well as their relative difference (Rel. diff.). Positive shear forces correspond to anterior forces at the cranial end of the intervertebral discs.

<table>
<thead>
<tr>
<th></th>
<th>Compression</th>
<th></th>
<th>Shear</th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th>Rel. diff. [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>ALC [N]</td>
<td>95% CI ALC</td>
<td>FLC [N]</td>
<td>95% CI FLC</td>
<td>Rel. diff. [%]</td>
<td>ALC [N]</td>
<td>95% CI ALC</td>
<td>FLC [N]</td>
<td>95% CI FLC</td>
<td>Rel. diff. [%]</td>
<td></td>
</tr>
<tr>
<td>0kg</td>
<td>L1-L2</td>
<td>456</td>
<td>393</td>
<td>518</td>
<td>406</td>
<td>361</td>
<td>451</td>
<td>-11</td>
<td>-93</td>
<td>-107</td>
<td>-78</td>
</tr>
<tr>
<td></td>
<td>L3-L4</td>
<td>448</td>
<td>394</td>
<td>502</td>
<td>401</td>
<td>357</td>
<td>445</td>
<td>-10</td>
<td>37</td>
<td>31</td>
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<tr>
<td></td>
<td>L4-L5</td>
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<td>568</td>
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<td>503</td>
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<td>L5-S1</td>
<td>494</td>
<td>425</td>
<td>564</td>
<td>451</td>
<td>402</td>
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<td>L1-L2</td>
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<td>L3-L4</td>
<td>1034</td>
<td>955</td>
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<td>1076</td>
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<tr>
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</tbody>
</table>

Non-overlapping 95% confidence intervals of the ALC and FLC results
6.5. Discussion

The aim of this study was to apply a novel approach of calculating the lumbar compression and shear forces during backpack carriage by combining physical and numerical modeling and to investigate the effects of load and lumbar curvature adjustment. The numerical simulations were conducted with a rigid body model [13], while the physical model provided the external forces acting on the upper body by the backpack. We evaluated three static scenarios, upright standing without load and with a backpack of 20kg and 40kg. The numerical simulations were conducted with adapted and fixed lumbar curvature. In case of the adapted lumbar curvature, upright MRI measurements provided in vivo lumbar curvature data as input for the rigid body model, whereas in case of the fixed lumbar curvature, a typical trunk forward lean angle was applied.

The study focused on the L4-L5 lumbar disc, as the cumulative compression and peak shear forces at the L4-L5 level are known risk factors for low back pain or injury [3]. Looking at Fig. 4, the increase in compression forces from 0kg to 20kg in the numerical model is much higher than the external compression force from the 20kg backpack in the dummy measurements (Fig. 3). The same is true for the shear forces and also for the increase of both types of forces from 20kg to 40kg. This difference was to be expected, due to increased muscular activity in vivo that is necessary to withstand and stabilize the external load of the backpack and it underlines the importance of the numerical model. The absolute values of the calculated L4-L5 compression and shear forces (Fig. 4) can be interpreted with existing recommended force limits for the lumbar spine, e.g. the recommended upper limit of 3400N for compression force by Waters et al. [25] and the 500N for shear force by McGill et al. [26] and Daynard et al. [27]. This implies the shear forces arising from the 40kg backpack condition could be critical at the L4-L5 and L5-S1 level. However, instead of defining a strict limit at discrete force values, it may be more helpful to reduce the compression and shear forces in the L4-L5 lumbar disc whenever possible, irrespective of the absolute values, in order to reduce the risk of low back pain or injury [3].

The effect of the load was significant for compression and shear forces at all lumbar levels, except for the shear forces at the L2-L3 level. The Helmert contrasts showed that the forces increased significantly with increasing loads. The effect of the lumbar
curvature adjustment was not significant for the compression forces at any level, but it was significant for the shear forces at the L4-L5, L3-L4, and L2-L3 level. Regardless of the fixed or adapted lumbar curvature, the shear forces below L3 were positive (Table 1), which corresponded to anterior forces at the cranial end of the intervertebral discs. Above L3, the shear forces were negative, corresponding to posterior forces at the cranial end of the intervertebral discs. The simulations with fixed lumbar curvature overestimated anterior shear forces and underestimated posterior shear forces for the 20kg and 40kg conditions (Table 1). At the L5-S1 level, this effect was not significant over all conditions, but for the 20kg condition, the non-overlapping 95% confidence intervals also indicate a significant difference. A possible explanation for the fact that lumbar curvature influenced the shear forces significantly, but not the compression forces, is that shear forces are primarily the result of gravity and therefore the lumbar curvature, or more specifically the orientation of the lumbar segments and the mass of all body parts above, had a significant effect. In contrast, the compression forces are mainly resulting from the muscular contractions, e.g. from the erector spinae muscles. The direction of these muscle forces in the lumbar spine is parallel to the longitudinal axis of the spinal segments. When the lumbar curvature changes, the muscles inevitably track these changes and the lumbar compression forces resulting from the muscular contractions stay the same, as long as the activation levels of these muscles do not change significantly. In summary, to predict realistic lumbar compression forces, subject-specific lumbar curvature data were not necessary for loads up to 40kg. Instead, the fixed lumbar curvature of the generic rigid body model was kept and for the 20kg and 40kg load condition combined with a typical forward lean. However, regarding shear forces, using subject-specific lumbar curvature data from upright MRI measurements as input for the rigid body model significantly altered lumbar joint force estimates. The above mentioned comparison of the absolute shear force values in this study with the recommended upper limit of 500N [26] already underlined the relevance of the shear forces. This means that for all cases, in which the total loading of the spine is of interest, subject-specific lumbar curvature data should be incorporated into spinal models.

The interaction effect of load and lumbar curvature adjustment was significant for the shear forces at the L3-L4, L2-L3, and L1-L2 level. The Helmert contrast further revealed that compared to 0kg, in the 20kg and 40kg condition the model predicted signifi-
cantly more anterior shear or less posterior shear with fixed than with adapted lumbar curvature. The difference between the 20kg and 40kg condition was not significant, but the fixed lumbar curvature results tended to overestimate the compression forces more in the 40kg (by 6-12%) than in the 20kg condition (by 2-6%). Therefore, the higher the load, the more the compression forces are overestimated with the fixed lumbar curvature.

Comparisons with literature values are not possible for the 20kg and 40kg condition, as there are currently no in vivo measurements of lumbar disc forces available during backpack carriage of comparable mass. However, a study by Rohlmann et al. [28] reported an average total force increase of 35N due to carriage of a 10kg backpack as measured by instrumented vertebral body replacements at the level of L1 (four subjects) and L3 (one subject). This average value resulted from a widespread range of five single measurements with a minimum of -8N (i.e. a decrease compared to standing without backpack) and a maximum of 96N [28]. When looking at the spinal forces predicted for the 0kg, 20kg and 40kg condition in this study (Table 1), a much higher spinal force increase might be hypothesized for a 10kg backpack condition. However, Rohlmann et al. [28] suggest that, despite high backpack masses increasing the spinal forces, backpacks with low mass of up to 10% of the subjects’ body weight may reduce the spinal forces, as the backpack counterbalances the forward bending moment of the upper body and allows for a reduced activity of the erector spinae muscles [22]. Therefore, a non-linear relationship between the increase of the spinal forces and the backpack mass must be assumed. Regarding the 0kg condition, a comparison of the absolute values presented in this study with reported values from literature was possible: Schultz et al. [19] reported a compression force at the L3-L4 level of 440N, which corresponds well with the value of 448N presented in this study for the simulations modeling an adapted lumbar curvature. The compression and shear forces on the other lumbar discs were in the same range as the values reported by Shirazi-Adl et al. [6] and El-Rich et al. [29], who used a finite element modeling approach to predict lumbar disc forces.

6.5.1. Limitations

Apart from the known limitations of the rigid body model already explained by Sente-ler et al. [13], the postural adaptations in our study are restricted to the sagittal plane.
Any possible effects of postural asymmetry, including scoliosis, are not included in this study and there is a need for future studies to investigate the loading of the lumbar discs and the risk of low back pain or injury during backpack carriage, including postural deviations in the frontal plane. Importantly, torsion has been previously identified as a risk factor for disc injury and degeneration [30, 31] and should be considered in future studies. In addition, all the investigated scenarios are static. Especially with regard to the peak shear forces, dynamic scenarios, e.g. walking or picking up the backpack, would be very interesting. However, the validation of the rigid body model used in our study is much weaker for dynamic scenarios [13] and the static nature of the MRI data is another obstacle. Lastly, the final subject sample consisted of seven subjects providing upright MRI measurements of three different load conditions. The original sample size of nine subjects was reduced to seven, as the corners of the vertebral bodies were not recognizable in the MRI images of two subjects.

6.6. Conclusions

This study demonstrated a novel approach of calculating the loads in the lumbar spine during backpack carriage by combining physical and numerical modeling. It is the first to use in vivo lumbar curvature data from upright MRI images with backpack carriage as input data for a rigid body model to calculate lumbar disc forces. To predict realistic lumbar compression forces, subject-specific lumbar curvature data were not necessary for loads up to 40kg. Instead, the fixed lumbar curvature of the generic rigid body model was kept and for the 20kg and 40kg load condition combined with a typical forward lean. However, regarding shear forces, using subject-specific lumbar curvature data from upright MRI measurements as input for the rigid body model significantly altered lumbar joint force estimates. Therefore, whenever the total loading of the spine is of interest, subject-specific lumbar curvature data should be incorporated into spinal models. The findings of this study add to the understanding of the relevance of lumbar curvature data in modeling load carriage scenarios. The presented approach of combining physical and numerical modeling can be applied to any existing numerical model that does not already contain a fully modeled interaction between a load carriage systems and the human body. Future studies may investigate the influence of different aspects, e.g. the backpack design, the hip belt tension, or the position of the center of mass of the backpack in order to decrease the lumbar loading and the risk of low back pain or injury during load carriage.
6.7. Acknowledgments

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6.8. References


7. Synthesis

Load carriage systems are used every day all over the world to transport material from one place to another. In the western countries, backpacks and vests are popular, as they allow free use of both hands during the load carriage. Typical fields of application include school children’s load carriage, infantry, or outdoor activities like trekking and hiking. However, carrying loads on the shoulders or on the back technically allows the application of very high loads [1, 2]. This can have at least two different negative consequences: the occurrence of discomfort, a feeling of soreness and numbness that is perceived during the load carriage and usually disappears after unloading; and the occurrence of pain and/or injury, which can appear either during or after the load carriage and usually does not disappear simply due to unloading.

Discomfort is less problematic from the health perspective, but it should not be neglected, as it is known to have a significant influence on user acceptance [3, 4]. It is therefore important for the users, who have a direct interest in minimizing the perceived discomfort, and also for the manufacturers, who can sell more load carriage systems if they offer reduced discomfort.

Regarding the occurrence of pain or injury due to load carriage, the most problematic injury risk from a long-term perspective is the risk of low back pain or injury. As opposed to other common injuries related to load carriage, the occurrence of pain resulting from an injury of an intervertebral disc often means that a full recovery is very difficult or even impossible: The first occurrence of low back pain increases the risk of a future occurrence of low back pain [5, 6]. A specific danger related to this kind of injury is that the process of injuring an intervertebral disc often remains unnoticed up to a certain stage. When pain finally signals a warning, the injury has already reached a critical state. In order to decrease the risk of low back pain or injury due to load carriage, known risk factors like the peak shear and the cumulative compression forces in the L4-L5 intervertebral disc must be minimized [7].

Extensive studies with in vivo measurements of the forces in the L4-L5 intervertebral disc are ethically impossible, so as an alternative, physical or numerical models need to be utilized to optimize load carriage systems regarding the risk of low back pain or injury. In contrast, when concerned about the discomfort, subject studies are very
common. However, for systematic studies and comparisons of load carriage systems, the use of evaluation tools based on models has several advantages compared to subject studies: Measurements with physical or numerical models enable lower costs, less bias, and higher test-retest reliability. Such an approach has been presented by the Ergonomics Research group at Queens University in Canada, who designed different models to assess different aspects of load carriage systems [8]. One of their models investigates discomfort, but it lacks a solid validation. Another one is built to measure the load distribution between the hip and the upper body, which could reveal interesting information about the spinal loading, but the exact forces inside the lumbar discs do not necessarily correlate with the external loading. In order to achieve substantial advancements in the field of discomfort during load carriage, the validation of a physical model to assess discomfort during load carriage was necessary. Due to a lack of appropriate data in literature, new in vivo data for a model validation had to be collected. To enable these measurements, an existing pressure sensor had to be assessed regarding its unknown suitability to measure the pressure distribution at the interface of load carriage systems and the human body. To be able to assess the risk of low back pain or injury during load carriage, the existing challenge of modeling the mechanical interaction between load carriage system and body had to be addressed. Additionally, the unknown relevance of precise lumbar curvature data in spinal models had to be investigated.

The aims of this work were therefore firstly to assess the reliability of Tekscan pressure measurements at the interface of load carriage system and body; secondly to apply this newly gained knowledge to a subject study, which was designed to measure the mechanical parameters that characterize the effect of a load carriage system on the body and to identify mechanical predictors of discomfort among these parameters; thirdly to use this novel in vivo data for a first validation of a physical model to assess discomfort during load carriage; fourthly, to present a novel approach of calculating the loads in the lumbar spine during load carriage by combing physical and numerical modeling and to evaluate the relevance of precise lumbar curvature data in spinal models.
7.1. Assessing the Reliability of Tekscan Pressure Measurements

Measuring the pressure distribution at the interface of a load carriage system and the human body is a technical challenge, as the currently available pressure sensors are not designed for measurements on a compliant and topographically complex surface. Additional challenges result from the temperature and humidity aspects of the human skin. Previous measurements on human subjects have been conducted with Tekscan sensors [9-11], but even in measurements on flat surfaces, these sensors are known to produce measurement errors up to 34% [12]. The aim of this subject study was to investigate and possibly improve the intra-subject test-retest reliability of Tekscan type 9801 sensors measuring the pressure distribution on the skin of subjects wearing load carriage systems.

The influence of several different factors on the intra-subject test-retest reliability was evaluated: the type of load carriage system, the activity of the subjects, the sensor location, and the data processing method. All measurements were conducted three times with each subject. The main outcomes revealed that the sensor location and the data processing method had the largest influence on the reliability. The hip and the shoulder are much more suitable for pressure measurements than the upper back or the chest. Concerning the optimal data processing, one of the tested data processing steps turned out to significantly improve the reliability of the data: The offset correction step significantly reduced the typical error as coefficient of determination, as well as the change in the mean. The concept behind this novel step of the data processing was to conduct base value measurements prior to the actual measurements. For these base value measurements, the sensors were mounted on the subjects but not yet loaded with the load carriage systems. By doing so, possible pressure artefacts due to the bending of the sensors were recorded. During the offset correction, the average signal of each sensor cell in the base value measurements was then subtracted from the actual measurement. However, even with the offset correction and the best-performing sensor location, the typical error as coefficient of determination was 24% or higher. A possible improvement for future measurements was expected by protecting the sensors against humidity entering the sensor cells, as this study revealed that humidity in the form of sweat could penetrate the sensor cells and result in an overestimation of the pressure. Subsequent measurements with three iterations on a single subject, comparing an unprotected sensor with a sensor
that was protected against humidity by a thin polyethylene layer, revealed a decrease of the typical error as coefficient of variation by a factor 1.6 for the protected sensor.

These results showed that the pressure measurement at the interface of the human body and a load carriage system remains a challenge. The novel findings of how to optimize the reliability of such pressure measurements were implemented in the following subject study. Specifically, the newly gained insights imply that careful consideration is required when choosing the sensor locations. Bony landmarks with a lot of motion involved, as it is the case e.g. in the upper back area with the underlying edge of the shoulder blade, are currently not suited for pressure measurements with Tekscan sensors. Additionally, for the more suitable sensor locations, the demonstrated offset correction step is highly recommended, while at the same time it is crucial to protect the sensors against humidity.

This study was limited to one pressure sensor type, but there was no specific alternative for this type of measurement. If future industrial development will contribute better sensor types, then the new sensors will have to be evaluated in a similar subject study to find out, whether any improvement in terms of the reliability will be observable.

7.2. Identifying Mechanical Predictors of Discomfort during Load Carriage

For the following study, the previous results meant that the sensor locations upper back and chest could be neglected. Luckily, for one of the most common load carriage systems, a backpack featuring shoulder straps and a hip belt, the most interesting interface areas, the shoulder and the hip, are also the sensor locations that provide the highest reliability. In order to prevent humidity from entering the sensor cells, all sensors were laser-welded into thin polyethylene bags, according to the novel findings of the Tekscan reliability study. The objective of this subject study was to provide the first in vivo measurements of parameters that characterize the effect of a load carriage system on the body and to identify mechanical predictors of discomfort during load carriage.

Twelve configurations of a commercially available backpack with a hip belt were investigated. The configurations varied in load and hip belt tension. As potential predic-
tors, the average and peak pressure in the shoulder and hip region, the strap force in the shoulder straps and the hip belt, and the relative motion between the body and the load carriage system in the shoulder and hip area were examined in a static and dynamic scenario separately. Static corresponded to upright standing and dynamic corresponded to treadmill walking. Discomfort in the shoulder and hip region, as well as overall discomfort were assessed using visual analogue scales in a questionnaire.

Static peak pressure turned out to be a significant predictor of discomfort, accounting for 85% of the variation in discomfort. Due to its measurement directly on the skin, this parameter is independent of the type of load carriage system and thus a generally valid mechanical predictor. Strap force can explain even more of the variation in discomfort, but this parameter is only valid for comparable load carriage systems of the type backpack with hip belt. A previous study by Stevenson et al. [11], investigating similar parameters, reported average pressure to be the best predictor, accounting for 31% of the variation in discomfort. The predictors found in this thesis are more than twice as powerful. A possible explanation for this difference could lie in the applied load carriage systems. The previous study used nine different load carriage systems [11]. One part of their unexplained variance could well have resulted from differences between the type and design of the load carriage systems. In contrast, the study presented in this thesis applied only one typical load carriage system and instead included twelve different configurations by varying load and hip belt tension. Hence, a lot of otherwise uncontrolled parameters like the angle of the shoulder straps or the center of mass of the load carriage system were kept constant. This successful identification of objectively measurable mechanical predictors was the main outcome of this subject study. It opens up much needed perspectives for the comparison of the discomfort of load carriage systems using physical models instead of subject studies.

As a secondary aspect, the differences in perception of pressure in different body regions, which have already been reported by Martin et al. [13], could be confirmed: The shoulder region was more than twice as sensitive as the hip region, regarding an increase in static peak pressure or strap force. According to these findings, user recommendations could be formulated: To decrease discomfort, the load should be shifted from the shoulders to the hip region, wherever possible. This has the benefi-
cial side effect of reducing the load on the lumbar spine, thus likely decreasing the risk of low back pain or injury.

A limiting factor of this study was the temperature dependence of the pressure sensors. If the subjects would have been asked to perform more strenuous activities, any increase of the skin temperature underneath the sensors would have also increased the temperature of the sensors themselves. As soon as the temperature of the sensors changes with regard to the state during calibration, the resulting absolute values are affected. This argument further promotes future measurements on a physical model, which would enable a controlled and constant temperature at the interface between the body and the pressure sensors.

7.3. Validating an Instrumented Dummy to Assess the Mechanical Aspects of Discomfort during Load Carriage

The novel identification of mechanical predictors of discomfort in the preceding study enabled a first detailed validation of an instrumented dummy to assess the discomfort during load carriage. The main objective of this study was divided into two tasks: first, to perform the actual validation by showing that the dummy can reproduce the mechanical parameters from the previous subject study; second, to evaluate the predictive power of the dummy in assessing the mechanical aspects of discomfort. The focus of this study was strictly on the mechanical aspects of discomfort, because the thermo-physiological aspects are well understood and existing physical models already allow the testing of thermo-physiological aspects of load carriage systems [14, 15].

The dummy features a male anatomy with a 3mm thick foam layer as a skin analogue. The static setup simulates upright standing with optional forward or backward leaning. To simulate walking or running, the dummy additionally performs a sinusoidal vertical motion with variable amplitude and frequency. In this study, all the parameters that were previously measured on the subjects were now measured on the dummy, while applying the same load carriage system with the same twelve configurations. For the first task, the correlations between the mechanical parameters from the subject study and the mechanical parameters from the dummy measurements were calculated. For the second task, the discomfort from the subject study served
as the dependent variable in multiple linear regression analyses, while the mechanical parameters from the dummy served as the independent variables. This revealed how much of the variation in the discomfort could be explained by the dummy measurements.

The correlation coefficients reported significant correlations of the mechanical parameters for all static measurements and also for all dynamic measurements in the hip region. In the shoulder region, for the dynamic average and peak pressure and the relative motion, the correlation was not significant, but the strap force made up for this by revealing a highly significant (p<0.001) correlation. The dummy presented in this study could thus be validated to accurately simulate the interaction between the human body and the load carriage system. Regarding the predictive power of the dummy, the regression yielded that static average pressure in the shoulder region and dynamic average pressure in the hip region accounted for at least 75% of the variation in discomfort. Using strap forces as predictors, even 79% or more of the variation in discomfort could be explained. The validity of the strap forces as predictors is limited to load carriage systems of the type backpack with hip belt, but the average pressure parameters are independent of the type of load carriage system.

The validated dummy to assess the mechanical aspects of discomfort during load carriage is the result of the first three studies of this thesis. It represents the first validated model that can replace subject studies in comparing load carriage systems regarding the mechanical aspects of discomfort. Its ability to measure a subjective perception with objective mechanical parameters, while being able to conduct hundreds of measurements at reduced cost and superior repeatability, opens up new possibilities to the field of load carriage system development. The load carriage system industry and the related research community can benefit by improving the design of load carriage systems, so that the users will benefit from decreased discomfort.

7.4. Assessing the Risk of Low Back Pain or Injury during Load Carriage

The instrumented dummy, which was already partly described in the preceding subject study, also featured the possibility to measure the load distribution between the pelvis and the upper body: A horizontal gap separated the dummy along the trans-
versal plane at the level of the umbilicus and two load cells measured the forces and moments in all three dimensions acting on these separate parts of the dummy. This load distribution information defines the external boundary conditions for the loading of the spine. However, the actual loads acting on the critical musculo-skeletal structures may also depend on other factors like the postural adaptations and the change in the muscular activity due to load carriage. An efficient way to take the muscular activity into account is to use numerical models. Such models also allow the calculation of the peak shear and the cumulative compression forces in the L4-L5 intervertebral disc, which are specific risk factors for low back pain or injury, as Kerr et al. [7] found out. However, in numerical models, introducing new load carriage systems and their interaction with the body is not efficient at all. The first goal of this study was therefore to bridge this gap by combining a physical model, the instrumented dummy, with a numerical model, which can simulate the load in the lumbar spine while accounting for possible postural and muscular adaptations.

The job of the physical model was to assess the interaction between the load carriage system and the upper body. The upper load cell inside the dummy provided this information. These external forces acting on the torso defined the external boundary conditions for the numerical model, a recently published, validated rigid body model [16], which is based on the open-source software OpenSim [17]. The rigid body model further allowed the input of lumbar posture data and then calculated the muscular recruitment based on minimizing the sum of the squared muscle activations. As a result, for static scenarios, the model accurately predicted the known risk factors that are associated with the risk of low back pain or injury during load carriage.

The biggest challenge in this approach was to gather accurate postural data of the lumbar spine for each load carriage system or configuration that was to be tested. A common, non-invasive way of measuring spinal postures is to use skin markers that are attached to the skin above the spinous processes or other anatomical landmarks. A motion capture system reports the three-dimensional position of the skin markers in space, which is then used to deduce the position of the skeletal structures inside the body. This procedure inevitably introduces inaccuracy, due to the relative motion of the skin with respect to the underlying bones [18, 19]. Considering this inaccuracy, it is important to understand how relevant the lumbar curvature adaptations due to load carriage are for the parameters of interest, the lumbar compression and shear
The second goal of this study was therefore to analyze the effect of precise lumbar curvature data in numerical modeling of load carriage. A subject study was conducted, which accurately assessed the lumbar posture of subjects wearing a load carriage system. Instead of using skin markers, the measurements were conducted in an upright MRI to maximize the accuracy while applying a minimal-invasive method. In return, the study had to be limited to analyzing static scenarios. The same backpack was applied as in the previous subject studies, this time with a total load of 20kg or 40kg. Additionally, upright standing without the backpack was included. The simulations were then conducted twice: once with subject-specific lumbar curvature that was adapted to the load condition and once with fixed lumbar curvature. This study was the first to use in vivo lumbar curvature data from upright MRI images with backpack carriage as input data for a rigid body model to calculate lumbar disc forces.

The results revealed that to predict realistic lumbar compression forces, subject-specific, adapted lumbar curvature data were not necessary. Instead, the fixed lumbar curvature of the generic rigid body model was kept and for the 20kg and 40kg load condition combined with a typical forward lean. However, regarding shear forces, using subject-specific, load-condition-adapted lumbar curvature data as input for the rigid body model significantly altered lumbar joint force estimates. Therefore, whenever the total loading of the spine is of interest, subject-specific lumbar curvature data should be incorporated into spinal models.

The presented novel approach combined physical and numerical modeling to predict the forces in the lumbar spine during load carriage and thus the risk of low back pain or injury. It provides a valuable and much needed extension to the field of load carriage system evaluation. Low back pain or injury is a serious health issue that affects every second adult [20]. As loads carried in load carriage systems tend to continuously increase [2], improvements in the design of systems will continue to be a constant need. Because in vivo measurement of the known risk factors for low back pain or injury due to load carriage are ethically impossible, the presented approach is a unique tool for the manufacturers and the research community. This study further revealed novel insights on the relevance of lumbar posture data in spinal models simulating load carriage.
7.5. Limitations and Outlook

In all subject studies conducted in this thesis, the subject population consisted of young healthy men. It is not clear to which extent the reported findings also apply to females and elderly people. While older people tend to be less typical users of load carriage systems, future research is needed to investigate possible differences between the genders regarding the discomfort and the risk of low back pain or injury during load carriage. Future studies must therefore also include physical models featuring female anatomies. Regardless of the gender, future research is needed to extend the assessment of discomfort from upright standing and walking to other postures and activities like crouching or climbing. The physical restriction exerted on the body by the load carriage system may play an important role. Regarding the pressure sensors used in the studies of this thesis, there is still room for technological improvements. This may offer new possibilities to increase the predictive power of models in the future. Another limitation of this study is the focus on the most common load carriage systems in today’s western culture. Additionally, the main area of application corresponded specifically to load carriage systems designed to carry high loads. Smaller loads usually do not pose a comparable challenge regarding the discomfort or the risk of injury. Further, the study that investigated the risk of low back pain or injury focused on static scenarios. A possible next step is to extend the presented approach to dynamic scenarios, which also calls for a numerical model that is validated for dynamic simulations. Lastly, the models presented in this thesis were intended to represent an average person among the young and healthy men. The anatomical and physiological variability in the human population make exact predictions for individual people impossible. Despite all validations, models can never fully replace human subjects. Models are extremely valuable to conduct hundreds of measurements in a reliable and controllable laboratory environment and at reduced cost. This provides freedom to investigate novel or unconventional concepts. Nevertheless, before a product is completely finalized, field studies will always be necessary to confirm the laboratory results, especially regarding the subjective perception of discomfort.
7.6. Conclusions

The novelties of this work include the following findings: a new data processing method that significantly improves the reliability of Tekscan pressure measurements on the body surface of subjects wearing load carriage systems by applying an offset correction; the necessity of protecting Tekscan pressure sensors against humidity in order to further improve the reliability; the first in vivo data of pressure, force and motion parameters that characterize the effect of a load carriage system on the human body; the identification of in vivo static peak pressure or strap forces as powerful ($R^2 > 0.85$) mechanical predictors of discomfort during load carriage; a first detailed validation of a physical model to assess discomfort during load carriage; a novel approach of assessing the risk of low back pain or injury during load carriage by combining physical and numerical modeling; the relevance of subject-specific lumbar curvature that is adapted to the conditions of load carriage as input data for spinal models. The presented approaches of assessing discomfort and risk of low back pain or injury during load carriage provide much-needed opportunities for future research and development of load carriage system types and designs.

7.7. References


Curriculum Vitae

Patrick Wettenschwiler

Born on 5th of May, 1984

Citizen of Rapperswil-Jona, SG, Switzerland

Education / Employment History

2012 – 2015  Dr. sc. ETH Zurich, Institute for Biomechanics, Zurich, Switzerland and Empa St. Gallen, St. Gallen, Switzerland

2011 – 2012  Empa St. Gallen, Switzerland: employment as project worker

2008 – 2010  MSc HMS ETH Zurich, Switzerland, Major in Biomechanics

2005 – 2008  BSc HMS ETH Zurich, Switzerland

Awards

2010  Willy-Studer-Award, ETH Zurich

Publications


**Conference Contributions**


2. Wettenschwiler PD, Stämpfli R, Lorenzetti S, Ferguson SJ, Rossi RM, Annaheim S. Body surface pressure as a predictor of mechanical discomfort during load carriage. *Annual Meeting of the Swiss Society of Biomedical Engineering; Zurich, Switzerland.* 2014

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Patrick Wettenschwiler