BIOMECHANICS OF BODY SUPPORT

A study of load distribution, shear, decubitus risk and form of the spine

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BIOMECHANICS OF BODY SUPPORT
A study of load distribution, shear, decubitus risk and form of the spine

BIOMECHANICA VAN LICHAAMSONDERSTEUNING
Een studie naar belastingverdeling, afschuifkracht, decubitus risico en de vorm van de wervelkolom

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Prof.Dr. M.W. Ribbe
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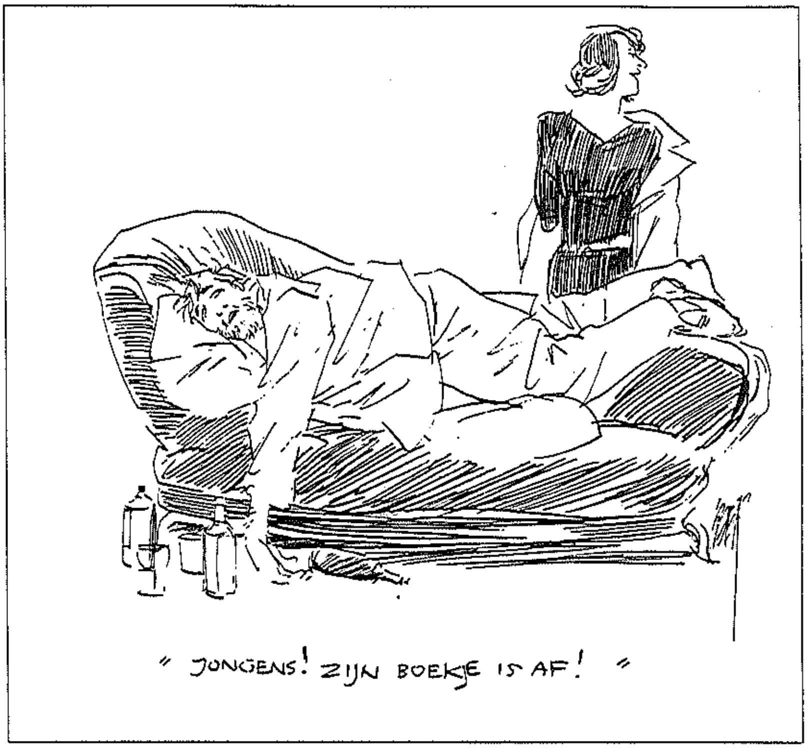
BIOMECHANICS OF BODY SUPPORT



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Voor Mique
Voor mijn ouders

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Goossens RHM, Snijders CJ. The curvature of the lumbar spine in sitting. Submitted.

Contents

Chapter 1

Introduction

Introduction

As human beings adopted the upright posture, sitting down became a convenient method of conserving bodily energy consumption[1]. Although e.g. Asian people adopt sitting postures by knee flexion while balancing on the feet and resting with the buttocks on the heels, such postures are not often seen in western society. It is assumed that the first chairs were rocks used to provide status in ancient times[2]. Since then the backrest has been added and the rocks replaced by soft and comfortable cushions. The homo sapiens had turned into homo sedens[3].

Nowadays chairs have become part of our lives, we sit at home, while travelling, at school and the tendency is toward more and more jobs being done in a sitting posture[4]. If it is estimated that we sleep 8 out of every 24 hours and that 80% of the remaining sixteen hours are spent in a sitting posture, this means that during a 72 year life-time, we spend 38 years sitting down.

Our constitution, however, is not evolved enough to adopt the unnatural sitting posture for prolonged periods of time. Therefore the sitting posture that is imposed by the workplace and in particular by the chair is often related to complaints in the neck, back, buttocks and legs[4]. Sometimes individual adaptations of the workplace are made in order to improve posture, although no agreement exists about the definition of good and bad postures. This is caused by the various scientific lines of approach in the studies on sitting problems. In research on the lower lumbar back, for example, it is advocated by several authors to tilt the seat surface forward in order to diminish a kyphotic curvature. However, a forward tilted seat surface may raise discomfort because of the shear force on the skin.

Anthropometrics form the most developed and well defined method of fitting a chair to a person; the body dimensions of individuals are used to define the dimensions of chairs. But even here problems soon arise, because the designer must take a specific sitting posture as the starting point for the matching of the dimensions of chair and person. In most office chairs this is the 90-90-90 posture, which means that the angle between thigh and shank, the angle between trunk and thigh and the angle between lower and upper arm are all 90° (see Figure 1-1). This posture is seldom seen in an office and only for short periods of time. Mandal[3] was one of the first to remark this.

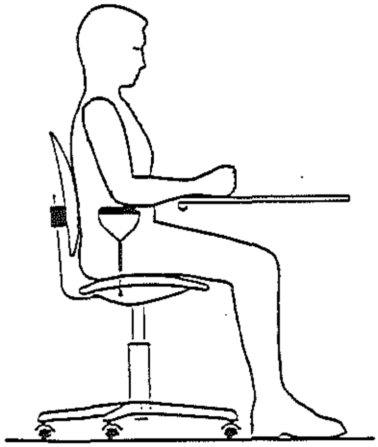

Figure 1-1. The 90-90-90 posture. The starting point for matching the dimensions of chair and person.

When collecting chair dimensions from different sources (appendix a), it is striking to see that the figures deviate considerably even when they are based on the 90-90-90 posture. This is caused by a difference in dimensions of individuals within populations (for example, as average Dutch persons are taller than Germans) and in the choice of the section of the population to be supported by the chair. Designers often use the rule of thumb that the tallest 5% of a population and the shortest 5% can be ignored. This is done to limit the range of adjustability and thus to save production costs. This resulted in foundation of pressure groups for the ignored 10% like the Foundation 'Stretched Out' in the Netherlands (In Dutch: Stichting Languit).

It is also often difficult to determine the backgrounds on which chair dimensions have been based. An example is the Dutch Standard, NEN1812, "Ergonomic requirements for office work chairs and office work desks, requirements for dimensions and design, measurement and test methods" of which not all data can be verified. The form of the backrest is said to be based on own research and experience, without giving details or references to these experiments which raises the

question as to whether the dimensions in the standard are adequate. The importance of this question increased with the introduction of new labour legislation in the Netherlands, many customers now demand that their new workplace is equipped with NEN1812-approved office chairs.
The shortcomings of using only anthropometics in chair design is notified in different studies[4-20], some of which use biomechanics to gain more insight into the forces that act on the seated subjects. Biomechanics can be defined as: the study of structure and function of biological systems by applying the methods of mechanics[21].
One field of biomechanics is related to the external load distribution on the different supporting surfaces. An example is the rule of thumb that sitting comfort is linked to the force that acts on the feet. When this force is larger than 1/3 of the total body weight, sitting becomes uncomfortable in the long run[4]. Stumbaum[5] measured all the forces that act on a seated subject with exception of shear force on the feet. He used his findings to design the tilting mechanism of an office chair. Snijders[6] introduced a biomechanical model that depicted the relation between the inclinations of seat and backrest in order to reduce the shear force on the seat.
Since the introduction of the first device that could measure pressure on deformable surfaces[7], numerous studies have been published on the pressure distribution on the seat[8-12].
Another field of biomechanics regards the internal body loads, like forces on joints, ligaments and muscles. The most famous study in this area is that of Keegan[13] who made a series of röntgenograms of the lower lumbar spine of healthy subjects. He was inspired by the observations of Åkerblom[2], that the lower lumbar spine flattened when seated. The most important finding in Keegan's study was that the vertebral bodies tilt, which enlarges the stress in the wedge-shaped intervertebral discs. After this observation, Nachemson *et al.*[14] and Andersson and co-workers[15-18] studied the pressure in the intervertebral discs of healthy subjects by inserting a needle. This resulted in the finding that the load on the intervertebral disc in unsupported sitting is 100% higher compared to standing (Figure 1-2).
With reference to these studies it is believed that a lordotic form of the lumbar spine is more healthy than a kyphotic posture. Mandal designed the famous forward tilting seat in the belief that this would lead to a

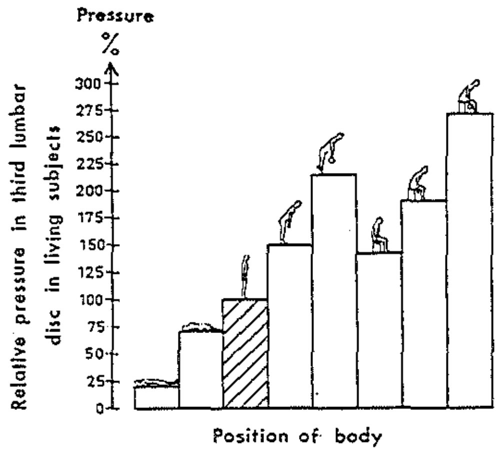

Figure 1-2. The load on the centre of the third lumbar disc in various body postures (source: Nachemson 1975)[22].

lordotic posture. In some studies the muscular activity is measured by means of electromyography (EMG) to study fatigue in sitting[15-20].
Although attention is given to biomechanics of body support for some time, there are still numerous aspects that are not studied. The reason for this is the absence of proper measuring devices. The department Biomedical Physics and Technology received grants (from the Ministry of Economic Affairs amongst others) to develop apparatus for the study of biomechanics of sitting. Subsequently the aim of this thesis was chosen as follows: to further quantify the consequence of body support on aspects that could not be measured up to the present, namely, the load distribution on all supporting surfaces, the distribution of the shear force on the seat, decubitus risk and the form of the spine in sitting.

These aspects were investigated in a series of studies that are reported in chapters 2 to 7.
Chapter 2 describes a new device that has been developed for the measurement of the load distribution on the body-supporting surfaces in different postures. With this apparatus a first series of measurements was performed on subjects sitting in bed.
Chapter 3 describes another device to measure the distribution of the shear force on the seat. Measurements of shear stress were performed on a wheelchair and on a hospital bed.
Chapter 4 describes the influence on skin oxygen tension when shear stress is applied to the skin. In this chapter the hypothesis that shear stress has influence on the decubitus risk was studied.
In chapter 5 the model of seated subjects by Snijders[6] was elaborated and validated with measurements on the device that was introduced in chapter 2. This chapter also deals with the question: What can be done to minimize the shear force on the seat?
In chapter 6 a new pressure sensor is introduced and a new method of evaluating dynamic pressure measurements was used. To evaluate wheelchair cushions in most studies, the pressure under the ischial tuberosities is measured in a static situation; wheelchair users, however, adopt different postures in time leading to different pressures, creating the need for a dynamic pressure sensor.
In chapter 7 the hypothesis of Mandal is tested, namely that a forward tilted seat produces a posture with a more lordotic form of the lower lumbar spine. This was already tested in a static situation, but sitting is a dynamic activity, therefore, an already existing device was adapted in order to measure the curvature of the spine continuously. This device is introduced in chapter 7 together with a series of measurements during a reading task.

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Chapter 2

A new instrument for the measurement of forces on beds and seats

Goossens RHM, Snijders CJ, Hoek van Dijke GA, den Ouden AH

Journal of Biomedical Engineering 1993; 15: 409-412.

A new instrument for the measurement of forces on beds and seats

Summary

Knowledge of the interaction of forces between persons and the bed in which they lay or the seat in which they are sitting, provides an insight into the loading of their muscles, bones and soft tissue.
To determine the total forces on the body supporting surfaces (backrest, seat pan, foot rest) resolved in components perpendicular and parallel to these surfaces a new instrument has been developed, with which the forces perpendicular and parallel to three different freely adjustable body supporting surfaces can be registered.
During the first measurements the forces on a bed were measured when a person sits in a bed with the backrest at an angle of 45° to the horizontal and the mattress horizontal. The measurements on a healthy population (mean mass = 77 kg, SD = 11 kg) showed an accuracy of ± 10 N. In this position the mean shear force on the seat pan was 97 N.

Introduction

Most readers will read this article in a sitting position. More than half the professions involve sitting, and the problems resulting from the use of seats have resulted in many different points of view in designing them. For example, in office chairs, two main ideas exist with regard to the design. In 1953 Keegan[1] prescribed the current office chair, which means a seat angle of about 5° tilted backward and a seat height based on shank height. Mandal[2], however, prescribed a forward-tilted seat for Danish school children and a seat height above popliteal level. Both researchers, however, departed from the same series of röntgenograms of the lumbosacral spine published by Keegan.
Knowledge of the interaction of forces between persons and the bed in which they lay or the seat on which they are sitting, provides an insight into the loading of their muscles, bones and soft tissue. Especially for

disabled people who use these provisions extensively, these forces may cause serious injuries (pressure sores)[3-7]. In addition the able-bodied people may have many difficulties resulting from the use of seats and beds, with the most common pain being low back pain[8].

Most research is devoted to pressure distributions on the seat pan[9-14]. Less research has been done on measuring forces on other body-supporting surfaces.

Stumbaum[15] measured nearly all forces acting on the supporting surfaces of seats. The missing force was the shear force on the feet, which he calculated out of equilibrium. With the aid of these measurements he developed a tiltable office chair. Eklund *et al.*[16] measured the vertical supporting force of a seated person on a floor-mounted force platform and calculated the biomechanical load on the lumbar spine at L3. Fleisher *et al.*[17] measured weight displacements on seats of chairs by means of a strain gauge and found individual load patterns while the subjects performed the same tasks. Gilsdorf *et al.*[18] used a force plate mounted on a wheelchair to measure shear force and normal force on the seat, and found that the wheelchair user should momentarily lean forward after a recline to reduce undesired forces.

The authors believe a twofold approach is necessary to analyze the forces acting on the users of seats and beds:

(i) Determine the total forces acting on the body-supporting surfaces (back rest, seat and footrest), resolved in components perpendicular and parallel to these surfaces.

(ii) Determine the local distribution of these forces over the surfaces.

The paper describes an instrument that can measure the total forces on the body supporting surfaces resolved in components perpendicular and parallel to these surfaces on different beds and seats.

Materials and methods

Two requirements underlie this instrument. Firstly, the instrument must be designed to install office chairs, wheelchairs and car drivers seats. To study the influence that the adjustment of a surface has on forces, it is

necessary to adjust it while the subject remains on the seat. Secondly, the forces must be registered for every possible installation with an accuracy of 2% (this is based on technical and economical considerations), resolved in components perpendicular and parallel to the surfaces and finally the point of application must be detectable.
For the rotations and translations of the body supporting areas of the different seats the guidelines for Dutch office chairs, NEN 1812[19], and the Dutch body dimensions, the DINED-table[20], were used as basic dimensions. By adding a surplus it was assumed that the instrument was suitable for the other seats and the disabled population. In addition to the different seats it also must be possible to install a bed. This has led to the requirements showed in Figure 2-1.

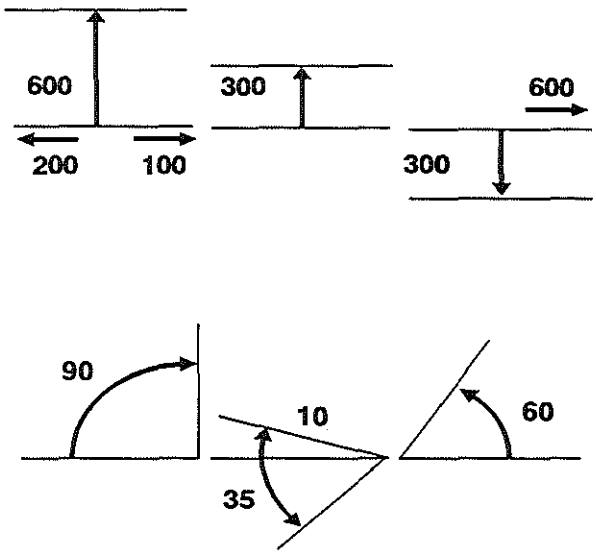

Figure 2-1. The rotations (°) and translations (mm) of the body-supporting areas, from an initial horizontal position (bed).

Requirements include the capacity to rotate the backrest over an angle of 90°, to rotate the seat forward 10° and backward over an angle of 35° and finally rotate the footrest over an angle of 60°. The foot support as well as the backrest must be adjustable in horizontal and vertical position, the seat only in the vertical position. Finally, it should be possible to rotate the seat and the backrest together over 25°. Mounted on the device are wooden plates to make it easy to assemble different chairs and beds.

To locate the point of application of the total force on the supporting surfaces, at least three different measuring positions per surface are needed. Therefore every surface is supported by three measuring-devices that work on the principle of strain gauges[21]. The whole construction has a total of 72 strain gauges, so that deformations perpendicular and parallel to the surface can be measured. The surfaces can be adjusted manually and by motor power and they are provided with a goniometer to measure the angle to the horizontal. The signals of all sensors are sent to a computer on which a software program is used to collect and process the data. The whole instrument is shown in Figure 2-2. The total

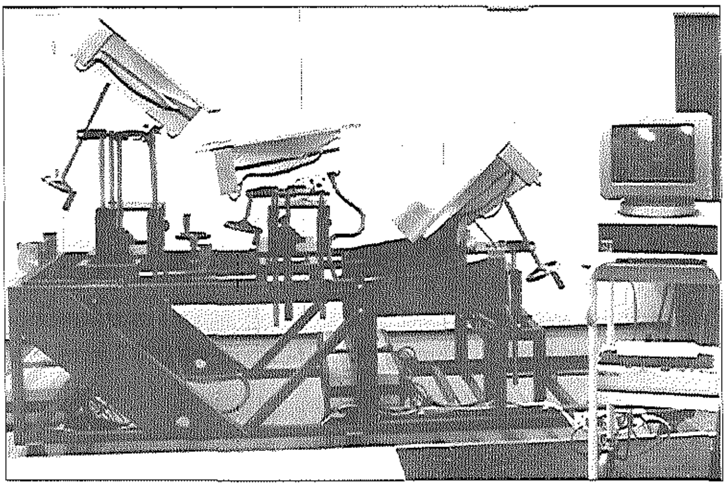

Figure 2-2. The instrument for measuring forces on beds and seats with graphical display of the data.

measuring chair needs to be calibrated only once before a measurement, namely when the original construction (e.g., an office chair) has been assembled. After that there is a software compensation for the weight of the surfaces under different angles.

The results can be displayed both graphically and numerically. In graphical display the total forces on the surfaces are displayed as vectors. When numerically displayed, the forces parallel and perpendicular to the surfaces, their horizontal and vertical components and the total horizontal and vertical forces are displayed by their mean and standard deviation.

The interaction of forces on the measuring device was tested. For this purpose a sit-in-bed position was installed. The backrest was adjusted to 45° to the horizontal, while the rest of the mattress remained horizontal; this is a position that is often found in hospitals. It was Reichel[22] who postulated that, when sitting like this, there would be a shear force on the seat pan and that it could be a factor in pressure sores. Indeed, Bennett *et al.*[23] found that the combination of pressure and shear was particularly effective in blood flow occlusion. In addition, the shear force causes the lying person to slide down slowly in his bed[24,25], which causes discomfort and extra work for the nurses.

The interaction of forces on ten healthy subjects (mean weight= 77 kg, SD= 11 kg) was measured on this bed configuration while the subjects all wore the same pair of cotton trousers from an operation suit and were supported on wooden plates. The forces were measured once in a series of 10 measurements, during which the subject was asked to put his hands on his lap and to sit still. The weight of the subjects was also measured on a digital balance and multiplied by 9.81 m/s^2 (accuracy $\pm$ 5 N). The paired test was designed to detect an absolute difference of 5 N between the means with a level of significance of 0.01 and a power of 0.95.

Results

Using the angles measured in the sit-in-bed position, the forces on the surfaces were resolved in their vertical and horizontal components and were added. In the unloaded situation all sensor outputs were registered for a period of 12 hours with intervals of two minutes during five days. These measurements showed that the output remained within $\pm$ 2 N.

With a subject on the instrument, the total horizontal force acting on the measuring-chair is expected to be zero (the subject is not accelerating horizontally) while the total vertical force should be equal to the weight of the subject. Table 2-1 shows the results of the measurements.

Table 2-1. An overview of the resulting total horizontal and vertical forces on the subjects as measured on the measuring-chair, and their weights measured on a digital balance (accuracy 5 N)

Digital balance		Measuring device	
Subject	Weight (N)	F_{vert} (N)	F_{hor} (N)
1	647.9	676.3	-2.3
2	665.1	663.9	-0.1
3	870.1	869.6	-0.3
4	765.2	769.1	0.3
5	563.1	564.1	0.0
6	737.7	741.3	-0.9
7	938.8	943.4	-1.6
8	845.6	850.7	-0.4
9	794.6	797.6	0.8
10	708.3	713.2	-1.4

A paired *t*-test was performed on the results of the weight and F_{vert} testing

$$H0:\ \mu_D = |\mu_{balance} - \mu_{device}| = 5\ \text{N}$$
$$H1:\ \mu_D = |\mu_{balance} - \mu_{device}| > 5\ \text{N}.$$

It showed that H0 should not be rejected ($\alpha=0.01$, Power$=0.92$) and this means that the accuracy of the total vertical forces on the device equals $\pm$ 10 N (2%).

A test on the total horizontal force was done, testing

H0: $|\mu F_{hor}| = 2$ N
H1: $|\mu F_{hor}| > 2$ N

Also in this test H0 was not rejected ($\alpha = 0.01$, Power=0.99).
Out of these tests it can be concluded that the total forces measured on the device, in this sit-in-bed position, have an accuracy of ± 10 N (2 %).

Discussion

Now that it is known that the forces have an accuracy of 2%, the components of the forces on the different surfaces can be determined. They are shown in Table 2-2 and Figure 2-3.

Table 2-2. Normal and shear forces during the sit-in-bed position. The shear force is defined positive when a subject slides out of the bed

	Mean (N)	SD (N)
$F_{Normal, foot}$	145	14
$F_{Shear, foot}$	41	10
$F_{Normal, seat}$	440	78
$F_{Shear, seat}$	97	17
$F_{Normal, back}$	221	35
$F_{Shear, back}$	25	9

The measurement of a mean shear force on the seat pan of 97 N with SD= 17 N support the hypothesis of Reichel[22]. The authors are currently conducting studies to determine the required seat tilt angle to obtain a lower shear force.

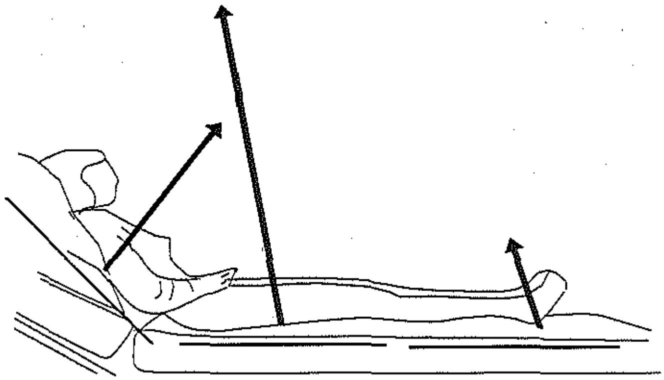

Figure 2-3. Graphical display of the mean forces acting on ten subjects while sitting in bed.

Conclusion

A measuring device has been developed and built, which registers the forces perpendicular and parallel to three different freely adjustable body supporting surfaces.
During the first measurements a posture simulating sit-in-bed, which is often found in hospitals, was installed in its original position. The measurements showed an accuracy of ± 10 N (2 %). With this sit-in-bed position a mean shear force on the seat pan of 97 N (SD = 17 N) was measured on ten healthy subjects.
In the future the use of this device may lead to criteria for the design of biomechanically sound seats and beds.

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Chapter 3

Shear stress measured on beds and wheelchairs

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Submitted

Shear stress measured on beds and wheelchairs

Summary

Local shear is understood to be one of the important risk factors for the development of pressure sores. There is a need for a small deformable sensor that can measure between skin and deformable materials without disturbing the shear phenomenon. In the present study a new sensor is introduced that approximates the design criteria. With a series of ten healthy young subjects validation experiments were performed.
It was demonstrated that with a forward tilted seat, the sum of the local shear forces between skin and sensormat is equal to the resultant shear force measured with a force plate. This result serves as a validation of the new sensor. The shear values recorded are 4.8 kPa in the longitudinal direction and 8.5 kPa in the transversal direction in sitting on a wheelchair and 5.6 kPa in the longitudinal direction and 3.1 kPa in the transversal direction on a mattress of a hospital bed. They remained below the limit of 9.9 kPa.

Introduction

All people that are forced to lie down or sit for longer periods, for example wheelchair users, have a risk to develop pressure sores.
Although the exact mechanisms behind pressure sores are not precisely known yet, researchers agree that mechanical load on the tissue is the main factor[1-7]. Some of these authors believe that paraplegia or denervat-ion is an extra factor in the occurrence of pressure sores. Daniel *et al.*[7] suggest that because of atrophy of tissue the padding around the bony prominences is reduced, resulting in higher interface pressures. Research on animals revealed an inverse relationship between intensity and duration of applied pressures. To explain this inverse relationship Reddy *et al.*[8] used a simple mathematical model on which preliminary analysis suggests that interstitial fluid flow may play an important role in ulcer formation. But all these authors agree that ischemic ulcers are due to

prolonged mechanical load, through which capillaries are closed and diffusion of oxygen and metabolites to the cells is hindered.
Tissue load in lying or sitting, can be influenced in two ways. Firstly, by changing the mutual positions of the body supporting surfaces. For example, when sitting in bed, the shear force acting on the buttocks can be influenced by changing the inclination angle of the upper legs[9]. Secondly, by changing the material and profile of the seat or the mattres. For example, sitting on a wooden surface produces higher pressures than sitting on a soft foam cushion. In most research the influence of the material on pressure is evaluated[10-16].
In 1958 it was Reichel[17] who started to focus the attention on shear force, defined as a force parallel to a surface, as an important component of mechanical load. Until now few articles have been published on this subject, because shear is extremely difficult to measure. Dinsdale[4] studied the effect of repeated pressure with and without shear in normal and paraplegic swine. He found that in those animals that received pressure and shear, ulceration occurred with lower pressure than in those animals that received only pressure.
Bennett *et al.*[18-20] studied the influence of shear on blood flow. He found that externally applied pressure was approximately twice as effective as shear in reducing pulsatile arteriolar blood flow. The combination of pressure and shear was found particularly effective in promoting blood flow occlusion in the palm of the hand.
The shear sensors used in previous studies, however, were only capable to be used on hard surfaces. Therefore the need was ascertained for a small, thin sensor that can be used on deformable materials.
In the present study a new sensor is introduced that has been developed to measure local shear stress, defined as a shear force per sensor area, acting on subjects in sitting and lying. The sensor is validated with the help of a force plate and its use in practice is demonstrated by measurements on a foldable wheelchair and a hospital bed.

Biomechanical aspects

Where can high shear forces be expected? Shear force can only exist when two surfaces are pressed against each other. The maximum shear force just before sliding occurs is defined by

$F_{Shear,max} = f \cdot F_N$

where f is friction coefficient and F_N is compressive or normal force. To hold body parts in position resultant compression and shear forces are active. These forces can be distributed over small or large areas, which involves respectively high and low stress. This implies that in regions where pressure is relatively high (a high compressive force), the shear stress can become high as well. The regions of relatively high pressure in sitting and lying are well documented in literature. The measurements of Garber *et al.*[12], Krouskop *et al.*[13], and Seymour and Lacefield[21] showed that during sitting, the area under the tuberosities has the highest pressure. In supine position the sacrum, heels, shoulders and caput receive the highest pressures[22]. The maximum pressure, which determines the maximum possible shear stress can be influenced by the choice of material. So, on a hard surface the highest shear stresses can be expected.

How can the magnitude of the shear stress be influenced? It can be expected that there is a relation between the local stress and the resultant shear force acting on a surface. When a low resultant shear force on a seat is wanted, Snijders[9] showed that the angle of the backrest and the seat must be coupled by means of a biomechanical model.

So it can be expected that the local shear stress can be affected by the choice of the material and by tilting of the seat and inclination of the backrest.

Materials and methods

When developing a shear sensor two aspects are of major importance.

Firstly, the surface of the sensor. When this is too smooth, correct magnitude of the shear force might not be detected, because the sensor slides over the surface.

Secondly, the dimensions of the shear sensor. The shear sensor always disturbs the continuity of the surface on which the measurements take place, therefore the sensor must be as flat as possible.
Shear can be detected by the deformation of a material under the application of shear force. An example of such a sensor can be found in Bennett *et al.*[19]. In the present study a sensor is created with the aid of two layers with electrodes separated by a layer of silicone rubber RTV 521, so that the shear force has a linear relationship with the displacement of the layers. The displacement has a linear relationship with the capacitance (Figure 3-1).

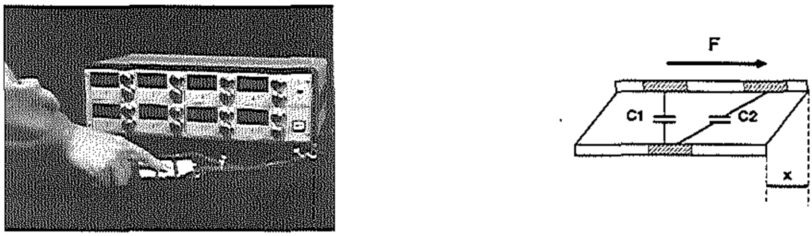

Figure 3-1. The principle behind the new sensor. A force (F) implies a certain displacement (x) causing a change in capacitance.

The prototype sensor was developed in close co-operation with Delft University of Technology and Erasmus University Rotterdam. The sensor is based on a capacitive principle by Heerens[23]. He used a principle that has been developed more than a century ago by Kelvin. In normal applications of capacitors Kelvins principle is not or not well implemented. That is why measurements by means of capacitors have the reputation of being very sensitive to displacement of the electrodes, but even more sensitive to disturbing influences of all conductors in the enviroment. Kelvin gave the solution to this problem by introducing a third electrode, the guard-ring. Heerens developed a theoretical background for the guard-ring principle and implemented it into different geometries. And thus very small changes in capacitance can be measured very accurately. In order to measure the displacement in two directions four capacitors are needed.

The sizes of the sensor are 27 x 15 x 3.5 mm, and thus its contact area 4.05 cm^2. Six of these sensors were glued on a thin slice of rubber (thickness 1 mm) to keep their relative position the same. In the appendix more details about the sensor and its specifications can be found.

I measurements to validate the sensor

In this test, shear on the seat was measured in two ways: resultant shear force with the aid of a force plate and local shear force with the new shear sensors. The resultant shear force was measured with a force plate that is part of a measuring chair[24]. The measuring chair has three force plates (backrest, seat, and foot support) on which a wide variety of seats and beds can be adjusted. Resulting shear and normal force can be measured on every body-supporting surface.
To measure local shear force the whole surface of the seat was devided into 4 rows and 8 columns. The sensor mat was put on all 32 positions. The shear stress measured by the sensor was converted into shear force by multiplication with the area of the sensor (4.05 cm^2). The shear force was now resolved in longitudinal and transversal direction of the seat. The sum of the measured 36 times 6 shear forces was compared with the resultant shear force as measured with the force plate, both in longitudinal direction. This procedure was repeated in 4 situations, with a 10° forward and a 10° backward tilted seat, with a hard surface (wood) and with a soft surface (foam, 5 cm thick). The backrest was not used during these tests.

II Shear measurements on the seat of a foldable wheelchair

Previous measurements[24] showed that a total shear force on the seat of a foldable wheelchair could become as high as 90 N. The high shear force occurs when the seat is horizontal. When the seat was tilted 8° backward, the shear force became smaller than 5N (with healthy subjects).
In the present wheelchair test local shear stress under the left buttock was studied in 4 situations:

- seat angle of 0°, backrest angle of 85°, no anti-decubitus cushion
- seat angle of 8°, backrest angle of 85°, no anti-decubitus cushion
- seat angle of 0°, backrest angle of 85°, anti-decubitus cushion based on gel
- seat angle of 8°, backrest angle of 85°, anti-decubitus cushion based on gel

Comparisons were made between a hard and a soft (gel) material and seat angles. In each situation subjects were measured five times. The maximum value of each measurement was used for statistics.

III Shear measurements on a hospital mattress

A study has shown that when a subject is sitting in bed with trunk angle 45° and legs horizontal, a shear force acts on the buttocks[25]. In this case study, the shear stress under the sacrum of the subject was measured locally. With a backrest angle of 45°, the angle of the legs was varied from 0° to 20° in steps of 5°. On each subject five measurements were done. The maximum value of each measurement was used for statistics.

For subsequent tests the sequence of the situations were randomized. Ten healthy subjects were measured (mass 68 SD= 14 kg, height 177 SD= 10 cm, age 24 SD= 3 years). Between every situation the subjects rose to allow adjustment of the measuring chair to be made. Before sitting down a correction for the offset was made. After sitting down, a period of two minutes was waiting followed before the data was collected. To ascertain equal coefficients, all subjects wore a cotton pair of trousers, as used in surgery. The measurements in test I were performed in the longitudinal direction only and in test II and III both in the transversal and longitudinal direction.
Statgraphics 5.0 was used for data analysis. In test I a paired t-test was performed on the results of the local and global measurements testing the hypothesis

H0: $\mu_{\text{sum of local shear}} = \mu_{\text{global shear}}$

H1: $\mu_{\text{sum of local shear}} \neq \mu_{\text{global shear}}$

In test II and III two factor analyses of variance were performed. The zero hypothesis (H0) stating that all means of the local shear force are equal, was tested against H1: at least one of the means is different, in all situations. A level of significance (α) of 0.05 was chosen.

Results

I measurements to validate the sensor

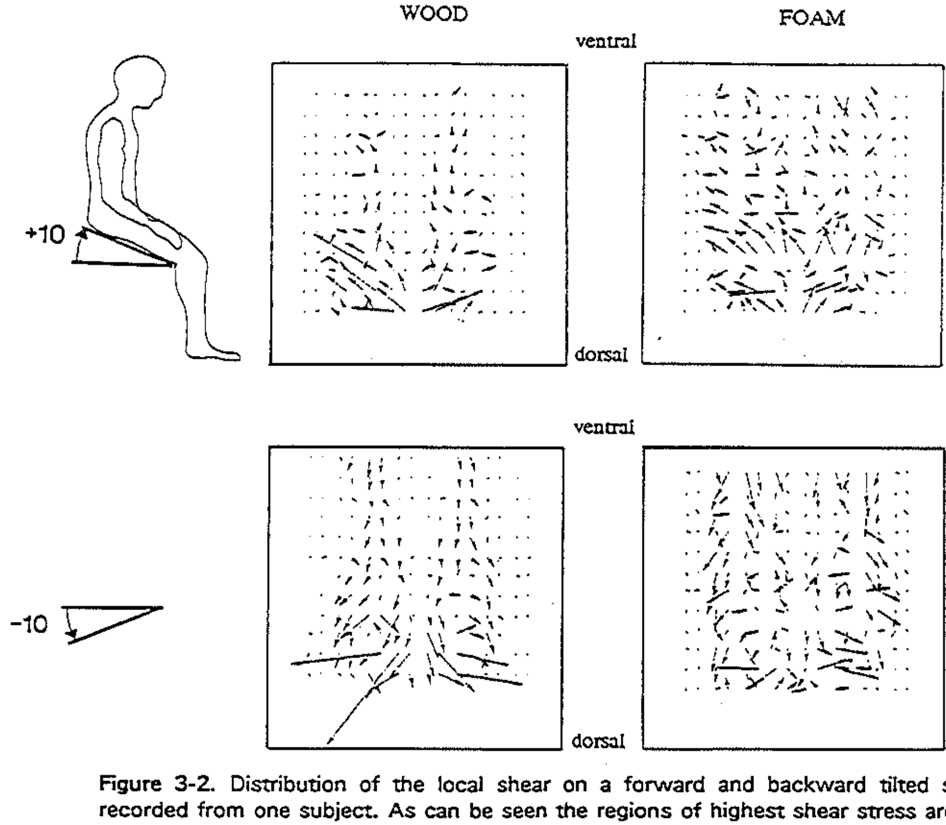

Figure 3-2. Distribution of the local shear on a forward and backward tilted seat recorded from one subject. As can be seen the regions of highest shear stress are at the ischial tuberosities.

A typical distribution of the shear stress in the four situations can be be seen in Figure 3-2. Table 3-1 shows the results of the sum of the local shear forces and the resultant shear force.

Table 3-1. Mean and SD (in brackets) of the local and resultant forces as measured on the seat from ten subjects

	Wood		Foam	
	Forward (10°) [N]	Backward (10°) [N]	Forward (10°) [N]	Backward (10°) [N]
Sum of local shear	85 (40)	-165 (52)	91 (22)	-155 (49)
Resultant shear	82 (15)	-112 (24)	87 (17)	-117 (23)

The H0 hypotheses had to be rejected in the backward tilted situation ($P=0.008$ on wood and $P=0.006$ on foam). This means that in the backward tilted situation, the sum of the local shear forces is not equal to the resultant shear force.

II Shear measurements on a foldable wheelchair

The results of the measurements are shown in Table 3-2.

Table 3-2. The mean of the maximum local shear force under the left tuberosity of ten subjects in kPa (SD in brackets)

	No cushion		Gel cushion	
	Longitudinal [kPa]	Transversal [kPa]	Longitudinal [kPa]	Transversal [kPa]
seat angle 0°	5.0 (1.8)	9.6 (3.9)	4.4 (1.2)	5.7 (1.7)
seat angle 8°	4.6 (1.3)	7.4 (2.5)	4.7 (2.6)	5.0 (1.7)

Analysis of variance was performed on these results. In the longitudinal direction the difference in shear stress was not significant. In transversal direction there was a significant difference of shear stress between wood and gel, with a seat angle of 0° ($P=0.01$) as well as a seat angle of 8° ($P=0.02$). In both situations the shear stress in the transversal direction was lower when the gel cushion was used.

III Shear measurements on a hospital mattress

The following results were measured, Table 3-3. Analysis of variance showed no significant difference for the different seat angles, neither in the longitudinal nor in the transversal direction.

Table 3-3. The mean and (SD in brackets) in kPa of the maximum local shear force on the sacrum of ten healthy subjects lying in a hospital bed with different seat angles

Seat angle	Longitudinal [kPa]	Transversal [kPa]
0°	5.5 (1.2)	3.1 (0.9)
5°	5.6 (1.3)	3.0 (1.0)
10°	5.8 (1.5)	3.1 (1.1)
15°	5.6 (1.6)	3.3 (1.5)
20°	5.3 (1.7)	3.0 (1.2)

Discussion

Measurement of shear without influencing the phenomenon is difficult. An ideal sensor should be flat, flexible and small. Although the sensor introduced in this study is relatively large and stiff, shear can be measured locally. When this sensor is used in studying pressure sores, a criterion is needed to interpret the data. The only criterion known from literature, is that local shear plays a role when it is above 9.9 kPa (75 mmHg)[18].

The design of the capacitor type sensor requires as dielectricum a material that has relatively large deformations under load. For that material silicone rubber (RTV 521) was chosen. This implies the disadvantage that after a load is applied, the deformation continues for a short period of time (creep). In this sensor 98% of the deformation is obtained after two minutes. Selection of rubber with less creep is in progress.

An attempt was made to validate the sensor by measuring local shear force together with the resultant shear force on a force plate. It was

expected that the sum of local shear forces equals the resultant shear force when all forces are in longitudinal direction. The results showed, that this is true both with a hard surface and soft covering. However, when the seat is tilted forward, the sum of the local shear forces is significant higher than the resultant shear.
Calculation of the total body mass of the subjects from the vertical force components of the force plates below the seat and the feet of the measuring chair showed that the resultant forces had an accuracy of 2%. From this it was concluded that the measurements of the resultant force were correct, although high as compared to the literature. Gilsdorf *et al.*[26] found a resultant shear force on the seat of 27 N on a hard surface and 50 N on a ROHO cushion. Stumbaum[27] found a global shear force on the seat varying from 0 to 60 N, depending on seat- and backrest angle. In both studies the subjects used a backrest, which must be the explanation for the lower shear forces in these studies.
The observation of a higher resultant shear force on the seat in the backward tilted situation can be attributed to a small backward shift of the subject with the exception of that part that is in contact with the shear sensors. It is likely that this results in the recording of local shear forces that are not necessary for equilibrium. This may not occur in the forward tilted situation, due to leg support that prevents the subject from sliding. This implicates that local shear measurements are only valid when such sliding does not occur. The use of a backrest probably prevents sliding by reducing the resultant shear force. In agreement with the expectation, the addition of local shear forces measured in the transversal direction showed no significant difference with zero (average 2 N, SD= 11 N).
In the second test shear stresses on soft and hard material were compared. The signifant larger shear stress, in the transversal direction on soft material can be explained by larger material deformation.
In the third test the relation between local shear stress and seat angle was studied. Previous measurements showed that the resultant shear force on the seat is decreasing with increasing seat angle backwards. This relation was also expected for the local shear stress. The results, however, showed no significant difference in local shear stress in the longitudinal direction for various seat angles on a foldable wheelchair (Table 3-2) or a bed (Table 3-3). A further study on one subject showed

that, when the subject remained seating during the change of the seat angle, the local shear stress changed slightly according to the expectation. Calculations based on the assumption that a resultant shear force is uniformly distributed over a seat surface revealed that, indeed, only slight changes could be expected (0.4 kPa), when the seat angle is changed. Local shear stress appears to be sensitive for body posture and movement. This came to light when the subject turned his head or lifted his arm. The shear stress changed to such an extent, that the effect of the change of the seat angle perished. Because in this study the subject stood up between every measurement, the small change in shear force due to the change in angle was not measurable. Still, the local shear forces presented in this study can have a practical value as an average for different situations; which can be compared with the permissable limit.

Local shear on a hard seat was measured by Bennett *et al.*[19]. He too found that the local shear stress is independant of the seat angle. The device used by Bennett *et al.* was based on strain gauges and the measurements took place on a geriatric and a healthy population. Shear was measured with a sensor with an area of 0.20 cm^2 (20 times smaller then the sensor used in this study) the position was 9 cm lateral of the seat center line and 3 cm lateral of the "ischial pressure locale". Measurements were done on a horizontal seat and a 20° tipped seat. Bennett *et al.* found that local shear stress with the geriatric population was almost 3 times as high as with the healthy population (average 2.6 kPa versus 0.8 kPa).

Conclusions

A sensor was developed for the measurement of local shear force. From tests on ten young healthy subjects the following can be concluded:

* The sum of the local shear forces between skin and sensormat is equal to the resultant shear force measured with a force plate, when the seat is tilted forward. This result serves as a validation of the new sensor.
* highest shear stress is at regions of highest pressure.

* a relation between local shear stress and seat angle can not be determined, possibly due to influences of body posture.
* in a standard foldable wheelchair the avarage local shear stress at the tuberosities equals 4.8 kPa in the longitudinal direction and 8.5 kPa in the transversal direction.
* when sitting in a hospital bed the avarage local shear stress at the sacrum equals 5.6 kPa in the longitudinal direction and 3.1 kPa in the transversal direction.
* local shear stress is highly affected by change in body posture, like head and arm movement.

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Appendix

The most important requirements of the shear sensor were as follows:

- thin, small and flexible in order to minimize bias
- shear sensitivity in two perpendicular directions
- range of at least 20 N per sensor area of 4 cm^2 (Based on 120 N shear measured in a bed on an estimated area of 1000 cm^2. With uniform shear distribution 0.5 N is expected on 4 cm^2. To deal with non-uniformity 0.5 N was multiplied by 40).
- hysteresis within 0.5 kPa
- no cross talk from pressure
- independance of temperature variations between 20 °C and 40 °C
- economic fabrication
- accuracy within 0.5 kPa

For calibration purposes, the shear sensor was put on a turntable with a weight on top of it. The table was put water-level and a shear force was

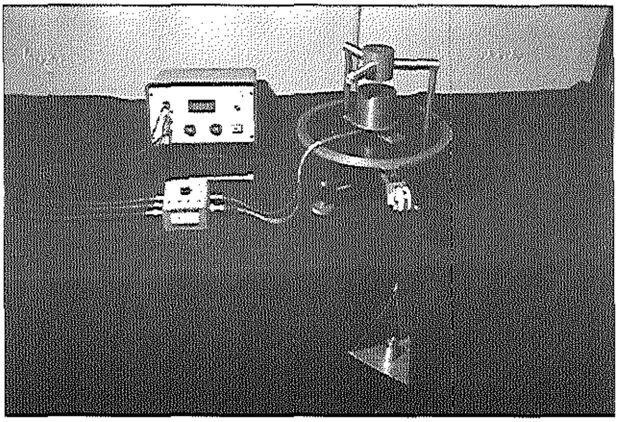

Figure 3-3. The measuring unit that was used to calibrate the sensor. One weight stands on top of the sensor. Another weight pulls by means of a string in longitudinal direction.

applied by pulling in longitudinal direction of the sensor by means of another weight, see Figure 3-3.
By rotating the table, the shear force in longitudinal direction of the sensor can be varied from 0 to a maximum value. After a load was applied to the sensor 3 minutes had to elapse before the rubber was fully deformed; this is a well known effect that is inherent for rubber. However within two minutes the value could be measured within the desired accuracy of 0.5 kPa. Figure 3-4 shows the output of the sensor as a function of the time.

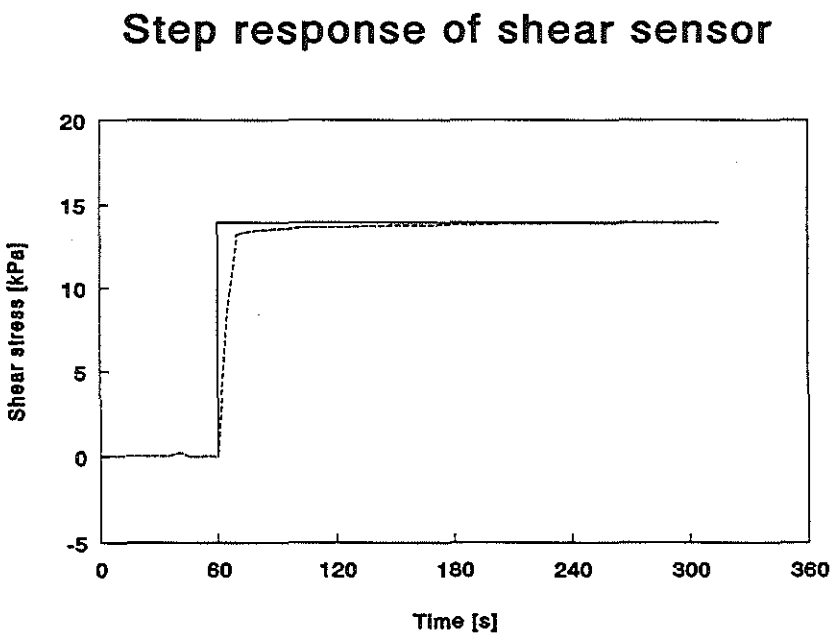

Figure 3-4. Output of the sensor as function of the time, when the sensor is loaded at t=60 s. The load is represented by the continuous line and the response of the sensor by the dotted line.

The turntable was rotated 90° in steps of 10° to load and unload the sensor. Everytime after application of the load a period of two minutes was waited. The curvature that was measured when loading and unloading can be seen in Figure 3-5. A least square fit is performed on the data in order to calculate the slope and intercept of the line. The maximum error was 0.3 kPa.

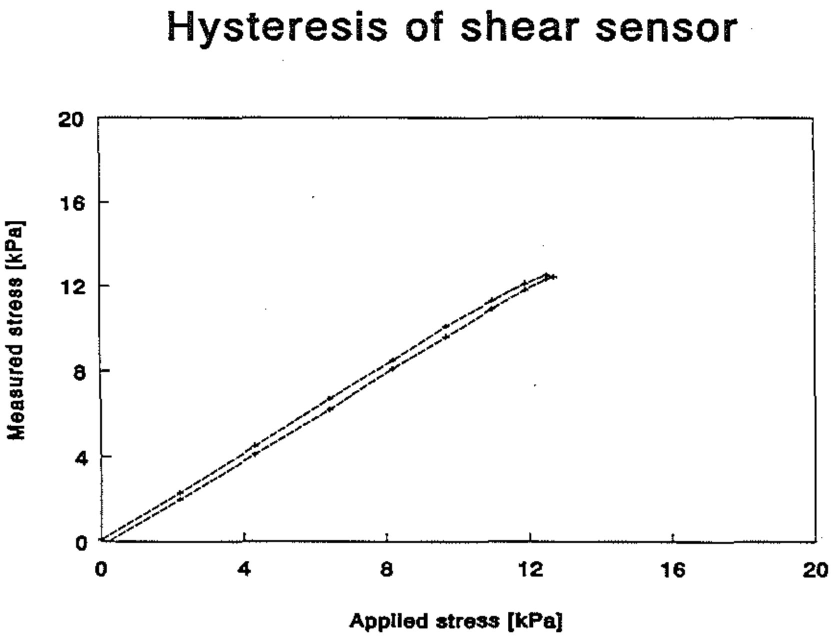

Figure 3-5. Loading and unloading curve of the sensor. Little hysteresis was found with a maximum of 0.3 kPa.

To test for temperature effects, the sensor was heated from 20°C to 40°C. In the unloaded situation the offset of the sensor changed with 2.9 kPa. However the sensitivity did not change significantly with these temperatures. When corrected for the offset, the load could be read well within the 0.5 kPa accuracy at both temperatures.
Finally the crosstalk from pressure on the sensor was tested. When the pressure was varied from 0 to 26.7 kPa, the output of the sensor remained within 0.5 kPa accuracy.
Summarized the following specifications were found.

Table 3-4. Specifications of the new sensor

Characteristic		Unit
maximum load	350	N
range	50	N
hysteresis	.3	kPa
resolution	2	mN
bandwidth	20	Hz
accuracy	0.5	kPa
cross sectional area	4.05	cm^2

Chapter 4

Influence of shear on skin oxygen tension

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Clinical Physiology 1993; 13: 000-000

Influence of shear on skin oxygen tension

Summary

Since the early studies on decubitus many authors expect significant influence of shear on decubitus risk. In the study described here skin oxygen tension at the sacrum is measured with combined pressure and shear stress on a young healthy population (mean age 25.5, SD= 3.4 years).

Cut-off pressure is defined as the level of external pressure exerted on the skin at which the skin oxygen tension is 1.3 kPa, at this level ischemia of the skin can be expected[1]. A mean cut-off pressure of 11.6 kPa is found when no shear stress was applied. A significant lower mean cut-off pressure of 8.7 kPa with a shear stress of 3.1 kPa.

No significant relationship was found between cut-off pressure and systolic blood pressure, diastolic blood pressure, skin thickness at the sacrum, percentage of fat and skin oxygen tension in the unloaded situation.

Introduction

Decubitus is caused by factors that are classified as intrinsic and extrinsic. The intrinsic factors are related to the patients clinical condition and both the nature of the illness and its severity are relevant. The extrinsic factors, that may be influenced directly, are concerned with pressure, shear, temperature and humidity.

All authors agree that the most important cause of decubitus is the mechanical load (pressure and shear) on the skin[2-5].

Although most authors agree that decubitus is due to prolonged tissue ischemia caused by the mechanical load through which the capillaries are closed and diffusion of oxygen and metabolites to the cells is hindered, other mechanisms are reported in literature.

Reddy *et al.*[6] studied the effects of external pressure on interstitial fluid dynamics using a simple mathematical model, concluding that squeezing of interstitial fluid may play an important role in ulcer formation.

Meijer[7] states that it is most likely that local blood circulation under influence of pressure is controlled also by regulatory mechanisms, which partly can be nervous. This is based on the observation of higher susceptibility in partly denervated rats by Manley and Darby[8].
When decubitus is taken to be caused by ischemia, the diffusion of oxygen to the skin is interesting. This diffusion can be measured in several ways. One possibility is the measurement of blood flow in the tissue. Using a photoplethysmograph Bennett *et al.*[9] studied blood flow occlusion in the palm of the hand as a function of mechanical load. In his thesis Meijer[7] mentioned the possible use of Laser Doppler flowmetry but rejected its applicability because of the small penetrability which is in the order of 1 mm.
Another possibility is measurement of the transient skin-temperature response as a measure for the blood-flow response[10].
Finally, Seiler and Stähelin[11] measured skin oxygen tension (pO_2) under increasing pressure exerted on skin tissue at bony prominences and muscle-padded areas, finding that at bony prominences skin oxygen tension fell rapidly under increasing pressure. The studies of Newson and co workers[12,13] and Bar[1] confirmed the existence of a cut-off pressure at different skin sites, which represents a pressure level at which complete ischemia occurs.
All studies on skin oxygen tension under load are confined to the influence of pressure alone, and the influence of both pressure and shear (the total mechanical load) is not evaluated. Recent studies (Goossens, chapter 3) showed that shear is present at sites at decubitus risk in beds and wheelchairs.
In this study the influence of pressure and shear stress on skin oxygen tension (pO_2) is measured using a special measuring unit, developed to apply both pressure and shear stress on the sacrum of young healthy subjects. Combinations of pressure and shear stress that are found in daily practice were exerted.

Materials and methods

The measurements were performed on a young healthy population. For every subject systolic and diastolic blood pressure and the percentage of body fat and the thickness of skin at the sacrum was measured to test on eventual correlations. In Table 4-1 the data for test subjects are given.

Table 4-1. The data of the tested young healthy population

Sex	Age [year]	Mass [kg]	Height [m]	Syst. blood pressure [mmHg] (kPa)	Diast. blood pressure [mmHg] (kPa)	Fat [%]	Skin sacrum [mm]
m	34	125	1.87	135 (18.0)	100 (13.3)	34	55
m	22	70	1.83	115 (15.3)	80 (10.7)	13	11
m	23	92	1.86	125 (16.7)	90 (12.0)	24	23
m	24	78	1.94	120 (16.0)	80 (10.7)	13	10
m	26	99	1.97	125 (16.7)	85 (11.3)	20	13
f	24	63	1.75	120 (16.0)	75 (10.0)	24	21
f	25	63	1.70	105 (14.0)	70 (9.3)	28	25
f	28	60	1.72	115 (15.3)	70 (9.3)	29	23
f	24	63	1.70	125 (16.7)	80 (10.7)	25	27
f	25	64	1.70	110 (14.7)	75 (10.0)	22	16

Blood pressure was measured using the auscultatory/sphygmomanometric method. A skinfold thickness measurement device was used (Servier, Zoetermeer, The Nederlands) to estimate the percentage of body fat using skinfold thickness measurements at four sites[14].
Skin oxygen tension was measured using the Transcutaneous Oxygen/Carbon Dioxide Monitoring System (TCM3) (Radiometer, Copenhagen, Denmark). A combined $tcpO_2/pCO_2$ electrode was used with a mass of 4 gram and a contact area with the skin of 63.6 mm^2.

The mechanical load on the skin was applied using two unsters (Figure 4-1), with an accuracy within 0.3% (Pesola, Switzerland). One unster (1) was used to apply pressure, by lifting partially an extra mass (3) which was attached to the electrode. This mass was beveled so that the weight only rested on the electrode surface. The other unster (2) was used to apply shear stress, by pulling a string in a direction parallel to the skin surface. The point of suspension of unster 1 was positioned at such a height that 99% of the applied shear force was conducted through the skin. With this construction almost linear displacement of the electrode is realized with skin deformations in a range of 2 cm.

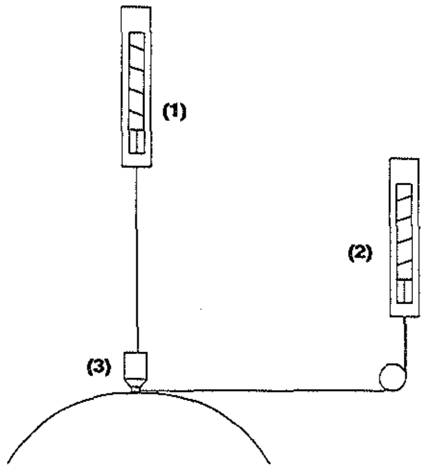

Figure 4-1. Measuring unit. Pressure and shear stress are applied to the skin using two unsters and a weight.

In relaxed prone position the electrode was placed on the sacrum in the middle of the triangle formed by both spina iliaca posterior superior and the cranial part of the nucleus glutealis. Before adjustment of the electrode, hair was removed and the skin was cleaned with alcohol. The hips of the subjects were bent (Figure 4-2) to relax the vertebral part of the iliopsoas muscle and to achieve a horizontal and flat sacrum.

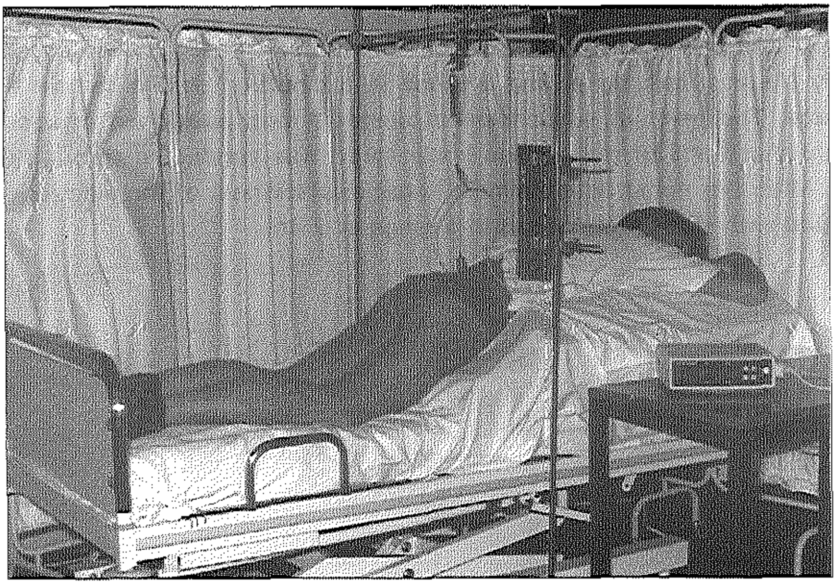

Figure 4-2. Measurement of oxygen tension of the skin at the sacrum loaded with pressure and shear stress. Relaxed prone position of subjects with the hips bent.

Pressure was applied at five levels, namely 0, 4.6, 7.7, 10.8 and 13.9 kPa and the heighest value was above the cut-off pressure found in literature[1]. The cut-off pressure is defined as the level of external pressure exerted on the skin at which the skin oxygen tension is 1.3 kPa. A study on 25 spinal cord injured subjects showed that, when the skin oxygen tension was below this level, ischaemic starvation had started underneath the tuberosities within two hours[1].

In the study described here this value was determined by linear interpolation of two near-by pressure levels. For example, when a pressure of 7.7 kPa gave a skin oxygen tension of 5.3 kPa and a pressure of 10.8 kPa gave a skin oxygen tension of 0.4 kPa, the cut-off pressure of 10.2 kPa was found.

The pressure was combined with two levels of shear stress, 0 and 3.1 kPa. For the highest shear load reference was made to previous measurements of local shear stress in beds and seats (Goossens, chapter 3). The different combinations of load situations were applied in random order. The friction coefficient between electrode and skin appeared to be high enough for the application of 3.1 kPa shear stress with 4.6 kPa pressure. The accuracy of the applied shear stress and pressure was 0.5 kPa.
The temperature of the electrode was chosen 44.5 °C to achieve local arterialization or maximal vasodilatation in the cutaneous tissue underneath the sensor. In that case, under normal circulatory conditions, the transcutaneous skin oxygen tension corresponds to the arterial oxygen pressure[11]. High temperature was used to dissolve the lipid structure of the dead cells in the epidermal layer. It was assumed that this was of minor influence to the response of the tissue to a mechanical load.
After the electrode was applied to the skin, a physiological stabilization period of 15 minutes was taken into account before loads were applied. A software program was written to compose graphs with pO_2 as a function of time. After the application of the load the skin oxygen was measured continuously and the measurement was done when a statio-nairy situation was reached.
To test the correlation between cut-off pressure and diastolic, systolic blood pressure, fat percentage, skin thickness at the sacrum and the pO_2 in the unloaded situation, a nonparametric correlation coefficient (Spearman rank correlation) was calculated using the software package Statgraphics 5.0. A 95% confidence interval for the mean skin oxygen tension of ten subjects at different levels of pressure and shear stress was calculated using the Students-t-distribution with 9 degrees of freedom.
Statgraphics 5.0 was also used to test the hypothesis:

$$\text{H0: cut-off}_{\text{shear stress}=0\text{ kPa}} = \text{cut-off}_{\text{shear stress}=3.1\text{ kPa}}$$
$$\text{H1: cut-off}_{\text{shear stress}=0\text{ kPa}} > \text{cut-off}_{\text{shear stress}=3.1\text{ kPa}}$$

in a paired t-test at alpha 0.05.

Results

Figure 4-3 illustrates the fall of skin oxygen tension after the application of a combination of pressure and shear stress. After a period of approximately 5 minutes a stationary situation was reached.

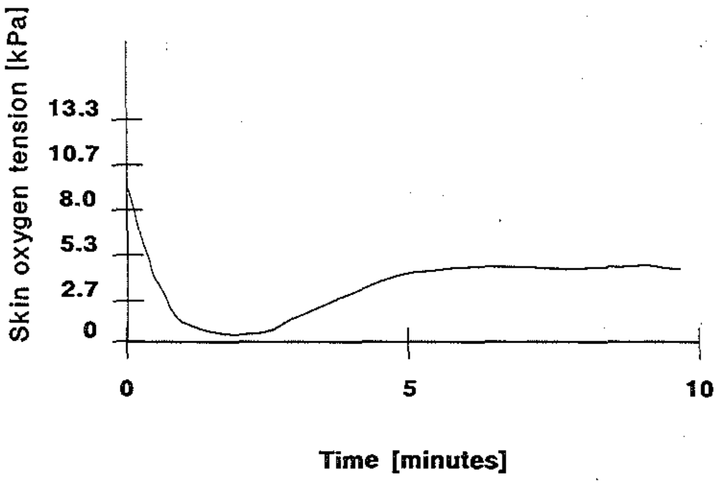

Figure 4-3. A typical example of skin oxygen tension as a function of time. The skin is loaded with 4.6 kPa pressure.

Figure 4-4 shows the means and the 95% confidence intervals of skin oxygen tension measured as a function of the applied pressure combined with no shear stress (continuous line) and shear stress of 3.1 kPa (dotted line). Clearly, the skin oxygen tension dropped with increasing pressure and was lower when shear stress was applied.
No significant relationship was found between the cut-off pressure and systolic blood pressure, diastolic blood pressure, skin thickness at the sacrum, percentage of fat and skin oxygen tension in the unloaded situation (Table 4-2).

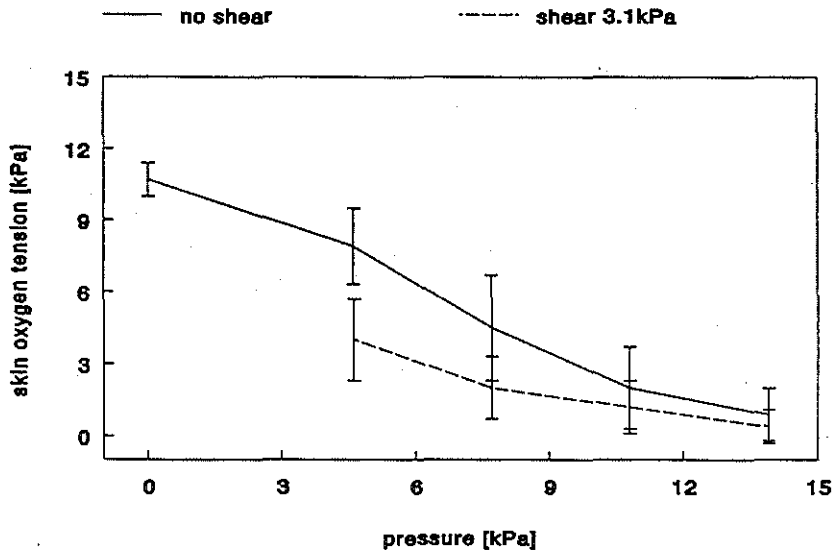

Figure 4-4. Skin oxygen tension as a function of pressure and shear stress, mean and 95% confidence intervals for ten healthy subjects.

Table 4-2. Coefficients of Spearman rank correlations between cut-off pressure and several variables (significance level)

	Cut-off pressure	
	Shear stress = 0 kPa	Shear stress = 3.1 kPa
systolic blood pressure	0.5989 (0.0724)	0.4768 (0.1526)
diastolic blood pressure	0.5433 (0.1031)	0.4242 (0.2032)
fat percentage	0.1890 (0.5707)	0.3180 (0.3400)
skin thickness at sacrum	0.4316 (0.1954)	0.5427 (0.1035)
pO_2 in unloaded situation	0.1155 (0.7290)	0.2012 (0.5461)

Finally, a paired t-test showed that the cut-off pressure was significantly higher ($P=0.0011$) in the absence of shear stress. A mean cut-off pressure of 11.6 kPa was measured when no shear stress is applied versus a mean cut-off pressure of 8.7 kPa when the shear stress was 3.1 kPa.

Discussion

Since the early studies on decubitus many authors expect significant influence of shear on decubitus risk. However, only Bennett *et al.*[9] performed measurements on this subject. With a photoplethysmograph they showed that occlusion of the blood flow in the palm of the hand occurred with a combination of pressure and sufficient shear stress (9.8 kPa). In more recent studies skin oxygen tension is used as a measure for decubitus risk[1,11,13] which revealed the existence of a cut-off pressure which causes ischemia.

In the study described here skin oxygen tension is measured with combined pressure and shear stress. A population of young healthy subjects was chosen, although a cut-off pressure found in this population may not be representative for the elderly and paraplegics. Compared to a healthy population Bennett *et al.*[15] found smaller blood flow volume rates in paraplegics and hospitalized geriatric patients. It is assumed that if an unfavorable effect of shear stress can be measured in healthy young subjects, the effect for the hospitalized geriatric and paraplegic population will be worse. Next to this method, photoelectric plethysmography and thermal clearance measurements probably may be used too. A comparative study of Newson and Rolfe[12] showed that skin surface oxygen measurements, thermal clearence and photoelectric plethysmography produced comparable cut-off pressures at the ischial tuberosities.

In the absence of shear stress the cut-off pressure of 11.6 kPa measured in the study described here appeared to be lower than the measured value of 13.3 kPa found by Bar[1] on a similar population (healthy, mean age 30 years) and the value of 15.2 kPa found by Newson *et al.*[13] (healthy population, mean age 26 years). However, Newson *et al.* defined the pressure at which pO_2 is 0 kPa as the cut-off pressure.

In the study described here, application of a shear stress of 3.1 kPa reduced the cut-off pressure significantly to 8.7 kPa. This level of shear stress proved to be far beneath the level found in literature[9].This is very important for the clinical situation because this shear stress level was measured with a new small deformable sensor that can measure between skin and deformable materials without disturbing the shear phenomenon (Goossens, chapter 3). A series of measurements were taken and an

average shear stress of 3.1 kPa was measured on the sacrum of ten healthy young subjects lying on the mattress of a hospital bed. On a wheelchair even higher shear stress was recorded.
In the clinical situation this measuring unit could be used in combination with pressure and shear stress measurements. First the load on a tissue site can be measured and this load can be applied to the tissue with this measuring unit to detect whether the tissue is deprived of oxygen.
It was hoped that a correlation would be found between on the one hand cut-off pressure and on the other systolic blood pressure, diastolic blood pressure, skin thickness at the sacrum, percentage of fat and skin oxygen tension in the unloaded situation. However, the only correlation that came close to significance ($p = 0.0724$) was between cut-off pressure and systolic blood flow.

Conclusions

Cut-off pressure is defined as the level of external pressure exerted on the skin at which the skin oxygen tension is 1.3 kPa[1]. This cut-off pressure is significantly reduced by shear stress of 3.1 kPa. A mean cut-off pressure of 11.6 kPa is found when no shear stress is applied versus a mean cut-off pressure of 8.7 kPa with a shear stress of 3.1 kPa.
No significant relationship was found between on the one hand cut-off pressure and on the other systolic blood pressure, diastolic blood pressure, skin thickness at the sacrum, percentage of fat and skin oxygen tension in the unloaded situation.

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Chapter 5

Design criteria for the reduction of shear forces in beds and seats

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Submitted

Design criteria for the reduction of shear forces in beds and seats

Summary

Both with respect to the aspect of pressure sores and of comfort, the inclination of backrest and seat are, amongst other factors, important design criteria. In this study the combination of seat and backrest inclination which reduces shear forces on the seat in passive seating forms the centre of attention. A biomechanical model is developed to predict these combinations and a new measurement apparatus was used for verification of the model on ten healthy subjects (age 24.4, SD= 2.1 years, height 1.77, SD= 0.08 m and mass 66.3, SD= 11 kg).

For chairs it was found that when little shear is accepted, a fixed inclination between seat and backrest can be chosen between 90° and 95°.

For beds a parabolic relationship was found between seat and backrest inclination with a maximum seat inclination of 20° at a backrest inclination of 50°. When lying with the knees bent to a position with equal inclination of upper and lower leg, the model predicts a shear force on the seat that shoves the person into the bed for every combination of seat- and backrest inclination.

Introduction

In modern society sitting takes up an increasing amount of time, both at home and at work. Grieco[1] investigated types of activity concerning sedentary work. It was concluded that there is a considerable shift to sedentary work in industrialized countries.

Given a certain function to be performed, a distinction can be made between active and passive sitting[2]. Passive sitting can be defined as a posture with minimum muscle activity, as opposed to active sitting in which more muscle activity is required.

Passive sitting can be realized by proper body-supporting surfaces. A poor chair design can cause body postures which can lead to unnecessary fatigue and even back and neck complaints, especially in prolonged sitting.

In order to obtain a posture with optimal body support, the chair must have at least armrests, a backrest and a seat. The mutual positions of the body supporting surfaces in combination with material properties and profiles influence body posture.

Body posture highly depends on the direction and distance of looking in combination with the work of the hands which is related to activity. In reading and writing with horizontal office desks the eyes force head and trunk in a forward bend position. When listening in a conference or a meeting and when using a visual display unit, the head and trunk are in a more erect posture. When sitting in a car seat or sitting in an easy chair, the head and trunk adopt an even more backward tilted position. In other words, different activities prescribe the position of the head and the angle of the backrest. Snijders[2] showed by means of a biomechanical model that the inclination of backrest and seat influences the shear force on the seat. The model predicts a relation between inclination of the backrest and seat in order to eliminate the shear force on the seat. In most of the hospital beds no attention is paid to this relationship. Goossens *et al.*[3] measured the forces on healthy subjects seated in bed with the backrest at an inclination of 45° with respect to the horizontal and the mattress in horizontal position. In this position the mean shear force on the seat pan was 97 N. This force explains the sliding of the patient on the mattress in the course of the day, leading to a sagged position with load on the sacrum. The combination of pressure and shear on the sacrum involves increased decubitus risk. Although the exact mechanisms behind pressure sores are not precisely known yet, researchers agree that mechanical load on the tissue is the main factor[4-9].

In 1958 it was Reichel[10] who started to focus the attention on shear force, defined as a force parallel to a surface, as an important component of mechanical load. Bennett *et al.*[11-13] studied the influence of shear on blood flow. They found that externally applied pressure was approximately twice as effective as shear in reducing pulsatile arteriolar blood flow.

In a recent study by the authors[14] it was found that shear stress of 3.1 kPa had significant influence on blood flow occlusion of the skin.

Both with respect to pressure sores and comfort, the inclination angles of backrest and seat are, amongst other factors, important design criteria. The present knowledge on the optimal mutual positions of body supporting surfaces is limited. Stumbaum[15] studied seated subjects in office chairs with the lower legs perpendicular to the floor. He only validated his biomechanical model on seats with backrest inclinations varying from 76° to 90°.
The aim of this study is to develop a biomechanical model which covers both seating and lying. In this study the combination of seat and backrest inclination which reduces shear forces in passive seating forms the centre of attention.
First a biomechanical model is developed which predicts these combinations, and a verification of the model is performed with a series of measurements on healthy subjects.

Biomechanical model

When the body adopts a sitting posture the weight of the body is distributed over the supporting surfaces. This distribution of weight depends on the mutual position of the surfaces. In the following a model is presented to calculate the forces in perpendicular and parallel components, with respect to the support surface.
The model is restricted to the sagittal plane and static situations and is an elaboration of the model introduced by Stumbaum[15]. In this model the seated human is divided into 4 links i.e. lower legs and feet, upper legs, pelvis and upper body. These links are connected by three joints (Figure 5-1).
With the assumption that these joints have no friction, the links can only apply push- and pull forces. Furthermore no muscle forces or ligament tensions are involved (passive seating) and no shear forces acting on the backrest.
In the model, the distributed shear and pressure on the supporting surfaces are represented by resultant forces. These resultant forces act on four major contact points, i.e. the feet, the ischial tuberosities, the crista iliaca and the thorax.

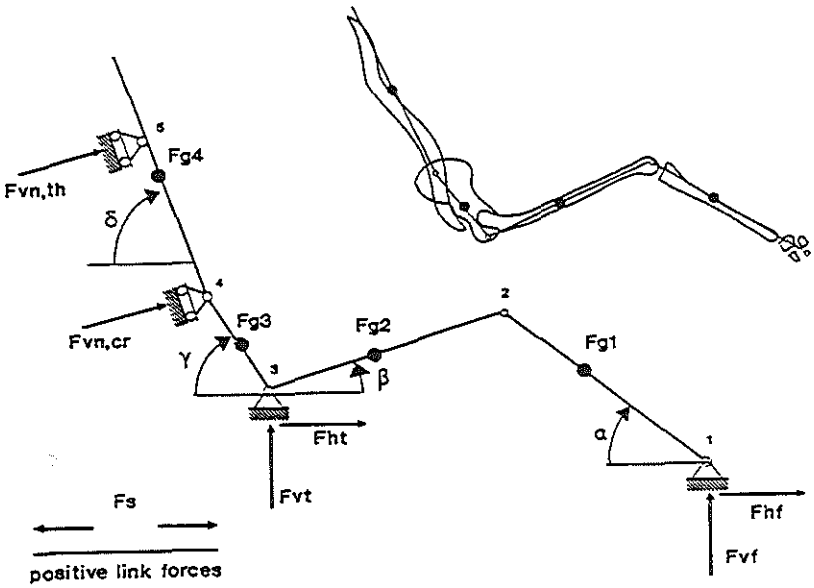

Figure 5-1. Four link model of the seated subject in the sagittal plane.

The following parameters are defined for the model.
inclinations

α	inclination of the lower legs
ß	inclination of the upper legs
γ	inclination of the pelvis
δ	inclination of the backrest

gravitational forces

F_{g1}	gravitational force on the lower leg
F_{g2}	gravitational force on the upper leg
F_{g3}	gravitational force on the pelvis
F_{g4}	gravitational force on the back

supporting forces

Fv_f	vertical reaction force on the feet
Fh_f	horizontal reaction force on the feet
Fv_t	vertical reaction force on the ischial tuberosities

Fh_t horizontal reaction force on the ischial tuberosities
$Fv_{n,cr}$ lower normal force on the crista iliaca
$Fv_{n,th}$ upper normal force on the thorax

link forces
F_{S1} link force in lower legs and feet
F_{S2} link force in upper legs
F_{S3} link force in pelvis
F_{S4} link force in upper body

Distance from centre of gravity to joint[15], Table 5-1.

Table 5-1. Distance of centre of gravity to joint

Centre of gravity	Joint number	Distance in % of link length
1	2	43
2	3	44
3	4	50
4	5	17

Gravitational forces as a percentage of total body weight[16], Table 5-2.

Table 5-2. Forces as a percentage of the total body weight

force	% of body weight
F_{g1}	13
F_{g2}	25
F_{g3}	22
F_{g4}	40

Using the distances from the centre of gravity to the joint numbers as mentioned in Table 5-1, the gravitational forces on the links can be transferred to the joints. An example follows for link 1 between joint 1 and 2. The distance from joint 2 to the point of application of the gravitational force is 43% of the total link length. When the total gravitational force is transferred to the joints, 43% of the force can be found in joint 1 (0.43 in equation 1) and 57% of the force in joint 2 (0.57 in equation 4). Now the following equations of equilibrium per joint are achieved.

joint 1:

$$0.43*F_{g1}+F_{S1}*\sin(\alpha)=F_{v,f} \tag{1}$$

$$-F_{S1}*\cos(\alpha)=F_{h,f} \tag{2}$$

joint 2:

$$F_{S1}*\cos(\alpha)-F_{S2}*\cos(\beta)=0 \tag{3}$$

$$F_{S1}*\sin(\alpha)+F_{S2}*\sin(\beta)=0.57*F_{g1}+0.44*F_{g2} \tag{4}$$

joint 3:

$$F_{S2}*\sin(\beta)+0.56*F_{g2}+0.50*F_{g3}+F_{S3}*\sin(\gamma)=F_{v,t} \tag{5}$$

$$F_{S2}*\cos(\beta)-F_{S3}*\cos(\gamma)=F_{h,t} \tag{6}$$

joint 4:

$$0.17*F_{g4}*\cos(\delta)+0.50*F_{g3}*\cos(\delta)+F_{S3}*\cos(90-\delta+\gamma)=F_{n,cr} \tag{7}$$

$$F_{g4}*\sin(\delta)+0.50*F_{g3}*\sin(\delta)-F_{S3}*\sin(90-\delta+\gamma)=0 \tag{8}$$

joint 5:

$$0.83 * F_{g4} * \cos(\delta) = F_{n,th} \tag{9}$$

Equations (3) and (4) are used to compute F_{S2}, while equation (8) is used to compute F_{S3}. These link forces are used in (5) and (6) to solve $F_{v,t}$ and $F_{h,t}$, which are the reaction forces on the seat.

$$F_{v,t} = \frac{(0.57 * F_{g1} + 0.44 * F_{g2}) * \sin(\beta)}{\cos(\beta) * \tan(\alpha) + \sin(\beta)} + (0.56 * F_{g2} + 0.50 * F_{g3}) + \frac{[F_{g4} * \sin(\delta) + 0.50 * F_{g3} * \sin(\delta)] * \sin(\gamma)}{\sin(90 - \delta + \gamma)} \tag{10}$$

$$F_{h,t} = \frac{(0.57 * F_{g1} + 0.44 * F_{g2}) * \cos(\beta)}{\cos(\beta) * \tan(\alpha) + \sin(\beta)} + \quad - \frac{[F_{g4} * \sin(\delta) + 0.50 * F_{g3} * \sin(\delta)] * \cos(\gamma)}{\sin(90 - \delta + \gamma)} \tag{11}$$

The starting point that there should be no shear force on the seat leads to equation 12. In this equation shear is defined positive when the subject slides out of the seat.

$$-F_{h,t} * \cos(\beta) - F_{v,t} * \sin(\beta) = 0 \tag{12}$$

The number of parameters in equations (10) and (11) are reduced by additional relations.

Stumbaum calculated that in sitting with the backrest in vertical position ($\delta = 90°$) a best fit of his model with measured supporting forces was found with a pelvis angle (γ) of 85°. For elaboration of this model to recumbent positions ($\gamma = 0°$) it is assumed that the pelvis is horizontal ($\delta = 0°$). Furthermore for intermediate positions an approximation is introduced by means of the linear relationship

$$\gamma = \frac{85}{90} * \delta = \frac{17}{18} * \delta \tag{13}$$

With (13) in (10) and (11) and the use of (12) a relation is derived between seat angle (ß) and backrest angle (δ) when no shear forces are allowed on the seat.

$$\frac{[F_{g4} * \sin(\delta) + 0.50 * F_{g3} * \sin(\delta)] * \cos(\frac{17}{18} * \delta + \beta)}{\cos(\frac{1}{18} * \delta)} - \frac{(0.57 * F_{g1} + 0.44 * F_{g2})}{\cos(\beta) * \tan(\alpha) + \sin(\beta)} - (0.56 * F_{g2} + 0.50 * F_{g3}) * \sin(\beta) = 0 \tag{14}$$

In this equation the first term is the contribution from F_{S3}, the second term is the contribution from F_{S2} and the third term the contribution from joint forces.

When $\alpha = 90°$ (lower legs perpendicular to the floor) the divisor of the second term becomes infinite and thus the second term becomes 0.

When the knees are stretched (α = -ß), the divisor of the second term becomes 0, which is beyond the validity of equation (14) because the term becomes infinite. Stretched knees are included in the model when it is assumed that there is no shear force on the heels and that the upper and lower leg form one link (Figure 5-2).

The relationship between ß and δ, which represents situations with stretched legs and without shear force on the seat, can be derived from equation 14 in the following way. The first term remains the same, the second term disappears, because $F_{S2} = 0$ and the joint forces in the third term change slightly. This leads to equation 15.

$$\frac{[F_{g4} * \sin(\delta) + 0.50 * F_{g3} * \sin(\delta)] * \cos(\frac{17}{18} * \delta + \beta)}{\cos(\frac{1}{18} * \delta)} - (F_{g1} + F_{g2} + 0.50 * F_{g3}) * \sin(\beta) = 0 \tag{15}$$

With equations 14 and 15, the following situations were calculated.

I Sitting with the lower legs at an inclination under 85° (equation 14) and variation of backrest between 70° and 86°. These values

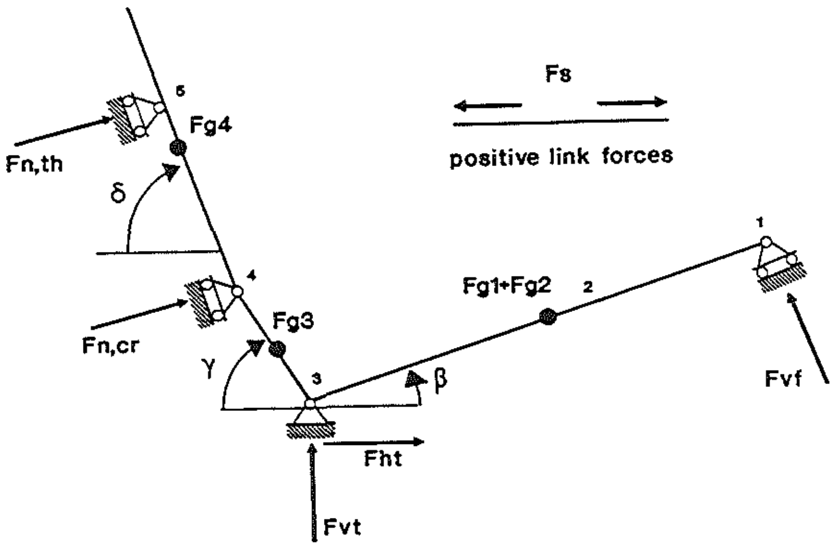

Figure 5-2. Model of the seated subject when the lower legs are stretched.

cover the range which is advised for office chairs and easy chairs[17,18].

II Lying with stretched knees (equation 15) and variation of backrest between 0° and 60°. The value of 60° is above the observed maximal inclinations of backrests in hospital beds.

The equations were solved using the software package EUREKA version 1.00. An error of 10^{-7} % of body weight was allowed.

Materials and methods

To determine the forces on the body-supporting surfaces (backrest, seat and footrest) in perpendicular and tangential direction, with respect to the support surface, the measuring chair was used. The measuring chair consists of three force plates by which a wide variety of chairs and beds can be simulated. The backrest can be rotated over an angle of 90°, the seat can be rotated forward 10° and backward 35° and the footrest can be rotated over an angle of 60°. Furthermore, the mutual positions of the surfaces can be adjusted in such a way that more than 90% of the Dutch population between 20 and 60 years of age, can sit or lie down on the device.

Mounted on the device are wooden plates to make it easy to assemble different chair- and bed-configurations. To locate the point of application of the total force on the supporting surfaces, three different measuring positions per surface are used. The measuring-devices on these positions work on the principle of strain gauges[19]. In addition, the surfaces are provided with a goniometer to measure the angle to the horizontal. Software is implemented to compensate for the mass of the wooden plates for different inclinations. It was shown in previous studies that the measured forces have an accuracy of 2%. A more detailed description of the device can be found in chapter 2 [3].

In the present study each supporting surface consists of a wooden plate without a cover. Before each series of measurements the measuring chair was calibrated.

In lying the backrest inclination was varied from 0° to 60° with steps of 5° in random order. In sitting the backrest inclination was varied from 70° to 86° with steps of 2° in random order. After adjustment of the backrest inclination, the edge of the seat was placed on a distance of 5 cm with respect to the hollow of the knee. Subjects were barefooted and the seat was given a height corresponding with an inclination of 85° of the lower legs. This was measured with a digital inclinometer (accuracy 0.5°).

After the subject had taken a seat, the inclination was changed until the shear on the seat in either direction was less than 1% of the total body weight. Although in theory it is possible to eliminate the shear force on the seat completely, it appears in practice that this is difficult to achieve due to slight changes in posture. Therefore in this study for shear force an arbitrary margin of 1% of the total body weight was applied. To get around the problem of friction, due to not coinciding centres of rotation in the subject and the chair during the alteration of the inclination, the subject was asked to stand up and sit down again after the alteration. The measurements were performed with the subject sitting still and relaxed with the arms on the lap. The subjective feeling of comfort was not questioned in this study. Ten subjects were used of which five were male.

All subjects were healthy and young (age 24.4, SD= 2.1 years) and their average height (1.77, SD= 0.08 m) and mass (66.3, SD= 11 kg) showed no significant difference with average height and mass of the Dutch population between 20-65 years (level of significance $\alpha=0.05$).

Statistics were performed using the software package Statgraphics version 5.0.

Results

I Model and measurements: seated subjects.

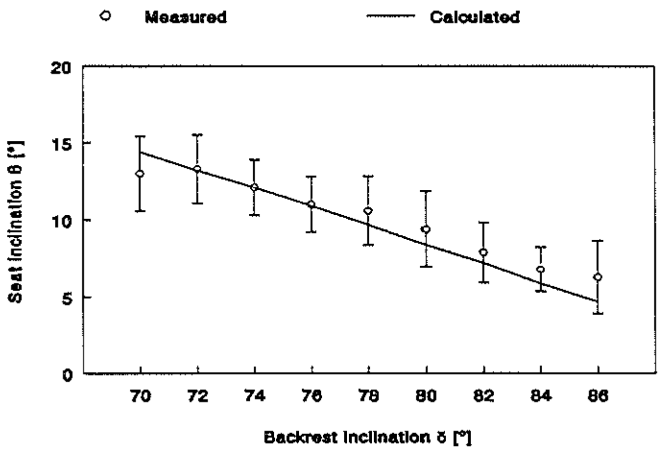

Figure 5-3. Seat and backrest inclinations during sitting that produced no shear force on the seat. The continuous line represents the combinations that are predicted by the biomechanical model. The circles with error bars represent the mean and 95% confidence intervals for the measured seat inclinations on healthy subjects.

In this series of calculations and measurements the combinations of seat- and backrest inclination were determined that produced less than 1% of body weight shear force on the seat during sitting.
Figure 5-3 shows the predicted and measured results. Predictions and actual values match well, the 95% confidence intervals for the mean include the predicted values for every backrest inclination. The predicted relationship between seat- and backrest inclination was found to be linear, i.e. $ß=-0.61*\delta+57.1°$ ($r^2=99.96$). (16)

II Model and measurements: lying subjects with stretched legs.

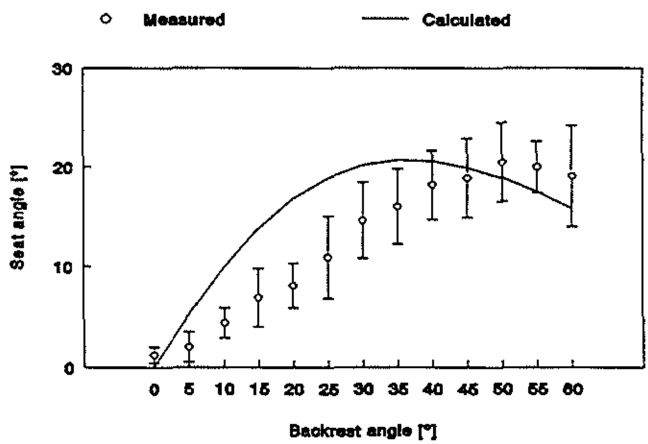

Figure 5-4. Seat and backrest inclinations during lying that produced no shear force on the seat. The continuous line represents the combinations that are predicted by the biomechanical model. The circles with error bars represent the mean and 95% confidence intervals for the measured seat inclinations on healthy subjects.

In this series of calculations and measurements the combinations of seat- and backrest inclination were determined that produced no shear force on the seat during lying.

Figure 5-4 shows the predictions and measured results. The measurements show that the relationship is parabolic and that the top of the graph corresponds with a maximum seat inclination of 20°. With a backrest inclination between 0° to 35° it can also be seen that a significant smaller seat inclination is required than is predicted by the model. The cause of the difference between model and measurements will be studied and commented in Discussion.

Discussion

It is known from literature[15] that the shear between body and seat depends on the inclination of the seat as well as the backrest. This observation is based on calculations and measurements only in sitting upright with vertical lower legs. In the present study a mathematical model is introduced which also covers positions in bed and any knee angle. The model simplifies the complex configuration of many body segments which raises the question whether model predictions and measurements of shear will correspond satisfactorily.
The measurements of body supporting surfaces on a series of healthy young subjects represented sitting and lying positions of daily life although the supporting surfaces were not covered with upholstering or cushions.
This model is restricted to the sagittal plane only and thus does not study the influence of postural changes in the frontal plane. In daily life, however, it is observed that postural changes in this plane do take place[20].
Very little is known about the influences of asymmetric postures on the supporting forces. Hobson[21] studied the influences of postural changes on the shear force acting on the buttock tissue in the plane of the seat. When the trunk was 15° laterally bend left and right, he found in the healthy population only little changes in shear force (5 N) compared to the symmetric position.
Two other studies were found in which the shear force on the seat was measured.
Stumbaum[15] accepted a small shear force on the seat, namely 5% of the total body weight. He studied healthy subjects when the backrest

inclination (δ) varied from 65° to 90°. The following equation that couples seat- (ß) and backrest (δ) inclinations was found:

$$ß = -\tfrac{1}{2} * \delta + 45°$$

This equation resembles equation 16 that was found in the present study.

Within the range from 70° to 86° of backrest inclination, the maximum difference in calculated seat inclinations (ß) in both equations, is only 4°. In order to reduce the shear force on the seat to zero, Hobson[21] extrapolated out of measured data that a backrest inclination (δ) of 55° should be combined with a seat inclination (ß) of 25°. This data is neither supported by the measurements in this study, nor by the biomechanical model (Figure 5-4).

For beds, the predicted seat inclinations are significantly higher than the measured seat inclinations, when the backrest inclination is between 0° and 35°. This can be explained by a deviation from the model in the pelvis region. With the backrest inclination at a range of 0° to 35°, the mass of the pelvis is not supported at the crista iliaca, but the pelvis rests completely on the tuberosities. This is supported by the measurement of higher normal forces on the seat than predicted by the model.

This can be modelled by transferring the mass of the pelvis (F_{g3}) to the ischial tuberosities (node 3). Formula 15 then changes into:

$$\frac{F_{g4} * \sin(\delta) * \cos(\frac{17}{18} * \delta + \beta)}{\cos(\frac{1}{18} * \delta)} - (F_{g1} + F_{g2} + F_{g3}) * \sin(\beta) = 0 \qquad (17)$$

With this model better predictions of seat inclinations are made (Table 5-3).

Table 5-3. Seat and backrest inclinations during lying that produced no shear force on the seat. The second column represent seat inclinations predicted by the *modified* biomechanical model. The third column represents the mean and SD for the measured seat inclinations on healthy subjects

Backrest inclination [°]	Seat inclination [°]	
	Model	Measured
0	0	1.2 (0.8)
5	3.3	2.0 (1.5)
10	6.3	4.4 (1.5)
15	9.1	6.9 (2.9)
20	11.4	8.1 (2.2)
25	13.0	10.9 (4.1)
30	14.8	14.6 (3.8)
35	15.0	16.0 (3.8)

Some remarks have to be made concerning the combinations of inclinations presented in this chapter. Although it is good to strive for a low shear force, other aspects can influence comfort as well. For example, when lying with stretched knees and a backrest inclination of 50°, shear force is eliminated with a seat inclination of 20°. This combination was tested in a hospital bed on patients that were bound to lie down during most part of the day. This induced discomfort, especially concerning the hamstrings. Bending of the knees might be a solution for this problem, provided that the patient is able to do so. When the knees are bent to a position with equal inclination of upper and lower leg ($\alpha = ß$), the model predicts shear force on the seat that shoves the person into the bed for every combination of seat- and backrest inclination.
For chairs the model prescribes a synchronization of seat- (ß) and backrest inclination (δ) according to the linear relation $ß = -0.61 * \delta + 57.1°$. This is practically insensitive (seat 1° steeper or flatter) for lower leg position between 80° and 90°. To implement this algorithm in chair constructions, a special mechanism is needed. However, when little shear is accepted, a fixed angle between seat- and backrest can be chosen between 90° and 95°.
From this study it can be concluded that the relation between seat- and backrest inclination can be predicted well by the developed model.

Application of the algorithms can be found in a wide range of chairs and beds, although the effect of soft cushions must be taken into consideration. This requires the measurement of local pressure and shear distribution.

Conclusion

A model was developed to cover the combinations of seat and backrest inclinations which reduces shear forces in passive seating. The model was validated with a series of measurements on young healthy subjects.
For chairs it was found that when little shear is accepted, a fixed inclination between seat and backrest can be chosen between 90° and 95°.
For beds a parabolic relationship was found between seat and backrest inclination with a maximum seat inclination of 20° at a backrest inclination of 50°. When lying with the knees bent to a position with equal inclination of upper and lower leg ($\alpha = \beta$), the model predicts shear force on the seat that shoves the person into the bed for every combination of seat- and backrest inclination.

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Chapter 6

Assessment of decubitus risk in a test circuit using different wheelchair cushions

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Assessment of decubitus risk in a test circuit using different wheelchair cushions

Summary

The risk of pressure sores is a common problem to wheelchair users. A great variety of wheelchair cushions are produced to reduce this risk. To evaluate wheelchair cushions in most studies the pressure under the ischial tuberosities is measured in a static situation with the subjects seated in one posture. Wheelchair users, however, do interact with their environment and during the day they will drive on different surfaces adopting different postures, leading to continuous pressure variations under the buttocks. Therefore in this study, the pressure under the buttocks of nine spinal cord injured subjects was measured continuously during wheelchair driving, while they were sitting on a foam cushion (thickness 5 cm) or a Jay Active™ cushion. The results of dynamic pressure measurements correlate with three classes of tissue response under the buttocks[1]: severe, moderate and mild.
The results from the data showed that a 5 cm thick foam cushion only produced severe responses, while a Jay Active™ cushion showed mild responses in most cases.

Introduction

The risk of pressure sores is a common problem to wheelchair users.
Many researchers studied pressure sores in animals. Pressure (and sometimes shear) was applied at different levels, for different periods of time.
Harman[2] studied the changes in skeletal muscle tissue of rabbits after subjecting the right leg to ischemia for several hours a day. He concluded that arterial occlusion initiates the process of ischemic necrosis, but that other causes must be sought to explain the perpetuation of the damage.
Kosiak[3] applied different levels of pressure for different periods of time on dogs' hind limbs. He found that ischemic ulcers in dogs were produced

by both high pressure applied for short durations and low pressure applied for long durations. In 1961 Kosiak[4] studied the effect of both constant and alternating localized pressure on normal and denervated muscle on rats. The data showed the marked susceptibility of tissue to relatively low constant pressure and a greater resistance to equal amounts of intermittent pressure. This was true for both normal and denervated tissue.

Dinsdale[5] studied the effect of repeated pressure with and without friction in normal and paraplegic swine. He found that in those animals which received pressure and friction, ulceration occurred with lower pressure than in those animals which received only pressure. Furthermore, he found that the results of blood flow determinations indicates that there is no significant difference in perfusion between applying pressure and pressure and friction. Therefore, he concluded that friction did not increase the production of ulcers by an ischemic mechanism.

Nola and Vistnes[6] used rats to determine the effects of pressure on skin and muscle. They found that a pressure-time regimen that always produced cutaneous ulceration over a bony pressure point, produced no ulceration in skin in a location where muscle separated skin and bone. But nevertheless significant areas of muscle necrosis in almost every case were present. Histologic studies of skin and muscle biopsies demonstrated epidermal breakdown, increased cellularity and muscle fiber necrosis when cutaneous ulceration occurred. They concluded that added thickness or bulk of tissue over a pressure point may lessen the amount of pressure applied per unit of surface area, and thus reduce the incidence of ulceration.

Manley and Darby[7] studied rats' hind foot after neuroectomy or tenotomy, after repetitive mechanical stress. It was concluded that normal levels of repetitive mechanical stress may cause plantar ulceration and that such ulceration occurs more readily in the denervated foot.

Finally, Daniel *et al.*[8,9] conducted experiments with healthy pigs on different pressure levels for varying durations. They concluded that the initial pathological changes occur in the muscle and subsequently progress upward toward the skin with increasing pressure or duration. When studying the role of paraplegia Daniel concluded that a pressure

sore could be produced in a shorter time in paraplegic animals than in normal animals given the same applied pressure.
Some of these authors believe that paraplegia or denervation is an extra factor in the occurrence of pressure sores. Daniel *et al.* think that because of atrophy of tissue the padding around the bony prominences is reduced, resulting in higher interface pressures. But all authors agree that ischemic ulcers are due to prolonged tissue ischemia caused by the mechanical load through which the capillaries are closed and diffusion of oxygen and metabolites to the cells is hindered. The study of Landis[10] showed that the mean pressure in the capillary vessels in human skin equals 32 mmHg (4.3 kPa), a value that is often found in literature as the pressure that may be applied to the skin without risk for the tissue.
To evaluate wheelchair cushions, in most studies the pressure under the buttocks is measured in a static situation with the subject seated in one posture[11-14]. Many different principles are used to measure pressure of which an overview may be found in the reviews of Holley *et al.*[15] and Crenshaw and Vistnes[16]. But wheelchair users do interact with their environment and during the day they will drive on different surfaces with different postures. It was already shown that in sitting a slight change in body posture introduces changes in the pressure distribution under the tuberosities[17,18]. In other words wheelchair use is a dynamic activity leading to continuous pressure variations under the buttocks[1,17].
Reswick and Rogers[19] found that different levels of pressure could be sustained for different periods of time. Therefore the dynamics of wheelchair use must have effect on the risk of pressure sores, leading to the conclusion that pressure should not be measured at one point in time but continuously. Therefore, Bar[1] monitored dynamic pressures beneath the ischial tuberosities of 25 spinal cord injured subjects. He calculated a quotient based on the pressure-time product above and below the threshold for anoxia. This quotient was used to make a distinction between three classes of tissue response.
In this study a new approach to evaluate the risk on pressure sores on wheelchair cushions is presented. The pressure under the buttocks of nine subjects with a spinal cord lesion was measured continuously while they were driving on six different surfaces, representing the ones that are met most during daily life.

To reduce this risk a great variety of wheelchair cushions are produced. One of these is the Jay Active^{TM}, a cushion that has three components: a urethane foam base, a Flolite pad and a cover. In this study the Jay Active^{TM} was compared to a foam cushion (thickness 5 cm) by means of dynamic pressure measurements during wheelchair driving.

Materials and Methods

A recording system was used that consisted of: the sensing elements for the pressure, a signal conditioning/input multiplexing module and a data recorder. The pressure sensor cells were flat, disc-shaped PVC cells that are commercially available (Talley Group Ltd, Hants, UK). The cells, with a contact area of 3 cm^2, were filled with a silicon fluid (Dow Corning 200, 1 cs.), and were attached to relative pressure sensors (Sensysm, SCX05DN) through PVC tubing (id: 1 mm od: 2 mm) of about 50 cm in length. Dynamic tests showed that, with proper liquid filling, the system had a bandwidth of 20 Hz. A similar system was developed by Barbenel and Sockalingham[20]. The electronic pressure sensors were attached to an ambulatory solid-state recording system RAMCORDER 1[21]; this is a device developed at our department that is capable of storing processed data from 8 input channels into digital memory. For the measurements, the number of inputs were expanded by using input multiplexing modules that contained signal conditioning amplifiers for the pressure sensors. The sensor signals were sampled with 8 Hz per channel.

The sensor cells were divided into two groups of seven cells, each group comprising one cell placed in the middle surrounded by six others (Figure 6-1). Before each measurement took place, the sensor cells were calibrated with the calibration equipment from the Oxford Pressure Monitor on three levels of pressure, namely 0, 100, 200 mmHg (0, 13.3, 26.7 kPa).

The sensor mats were positioned on a test wheelchair under the left and right buttocks of the subject. This was checked visually before the measurements started. Before beginning the test circuit, to be driven, the subject was seated for one hour on the cushion, because of other tests that took place on the subjects. Two cushions were used, a flat foam cushion (thickness 5 cm) and a Jay Active^{TM} cushion. A Jay Active^{TM}

cushion is only one of the many pressure relief cushions that are on the market. This particular cushion was used because the rehabilitation center we co-operated with prescribed it in the majority of cases.

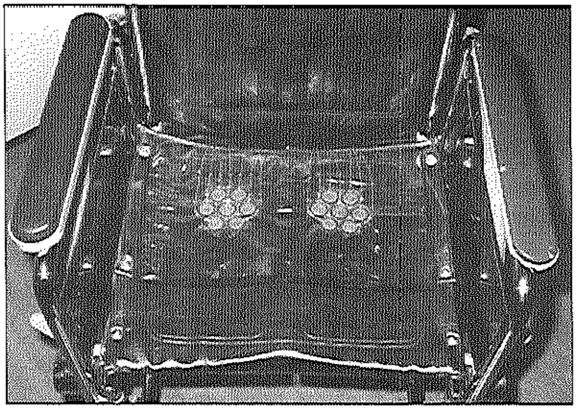

Figure 6-1. The sensor mats were positioned on a wheelchair under the left and right buttock of the wheelchair user. Because of the use of the RAMCORDER 1, a small portable data recorder, the subject could drive the wheelchair without wire connections while the measurements took place.

Each of the subjects completed six different conditions, for a period of three minutes per condition during which the pressure measurements took place. After the completion of each condition the subject rested for 12 minutes. The conditions were chosen to represent the surfaces that are being met during daily life. They were linoleum, a pavement, a low slope (5%), a curb of 5 cm height, a high slope (8%) and a coconut mat (Figure 6-2). Each condition was driven in this order by each subject. The subjects were all male and had a lesion at different heights, varying from C7 to Th12, average age 24.8 year, SD= 5 year.
The data were processed using the level of pressure applied on the buttocks above which no oxygen is available to the cells of the respective tissue. A threshold of 100 mmHg (13.3 kPa) was used in the case of young spinal cord injured subjects (lesion varying from L1-C2, average age 27.5, SD= 8 year)[1]. Above this pressure level there is an oxygen debt (D) and below this level an oxygen recovery (R) of the tissue cells occurs (Figure 6-3).

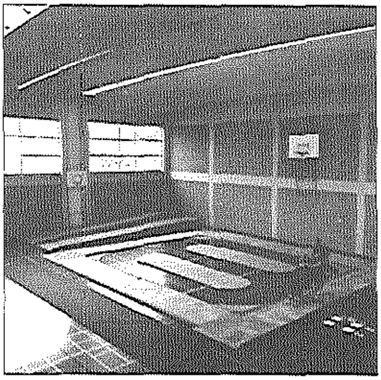

Figure 6-2. Six conditions representing surfaces that are being met during daily life by a wheelchair user.

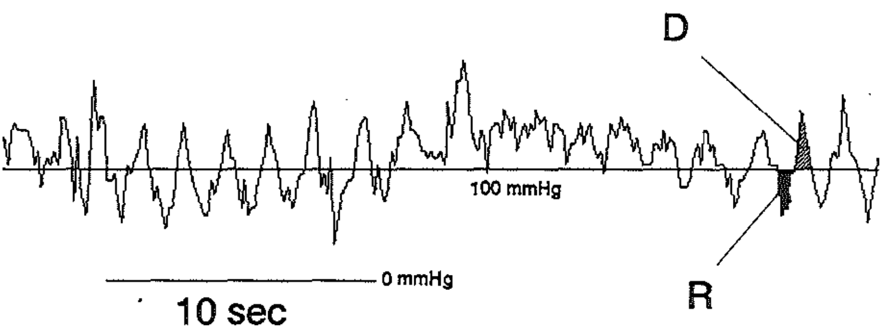

Figure 6-3. A pressure signal with an example of the intervals where oxygen debt (D) and oxygen recovery (R) of the tissue cells occurs.

From the data the quotient R/D was calculated.

$$\frac{R}{D} = \frac{\textit{area of pressure-time when measurements are below } 100 \textit{ mmHg}}{\textit{area of pressure-time when measurements are above } 100 \textit{ mmHg}}$$

Three different classes of skin response, based on redness and skin temperature, were found to correlate with this quotient (see Table 6-1).

Table 6-1. The relationship between the skin response under the buttocks of 25 spinal cord injured subjects and the ratio of oxygen recovery and debt of the tissue, found by Bar[1]

Response	R/D Ratio	Description
Mild	R/D>5	No marked erythema was observed but a mild flush of the skin sometimes occurred which faded within a few minutes.
Moderate	1<R/D<5	Marked erythema did not persist for more than 12 minutes and erythema generally faded within 20-30 minutes.
Severe	R/D<1	Concomitant with the appearance of an intense red flush of the skin and subsequent to its dissapearance was a sharp rise in temperature.

Results

The pressure on 14 sensor cells was measured per condition per subject. This means that about 22000 samples per subject per condition were recorded. These data were reduced by taking the mean value of each cell. This reduced the data to 14 mean pressure values. Further reduction was obtained by taking the maximum value from these 14 values only. So, one maximum mean pressure remains per subject per condition. The same kind of data reduction was done on the R/D quotient: only the minimum quotient of R/D per subject per condition was used for further analysis. The results of the dynamic pressure measurements on both cushions are presented in two ways. Firstly, the maximum mean pressures of subjects and conditions are given and a two-way analysis of variance is done on these results. Secondly, the minimum quotient of recovery and debt (R/D) is presented and the class of tissue response that this quotient belongs to. The maximum mean value of pressure is presented in Table 6-2.

Table 6-2. Maximum mean pressure acting on the buttocks of nine subjects whilst driving over six different conditions, for three minutes per condition (lin = linoleum, pvmt = pavement, low = low slope, curb = curb of 5 cm height, high = high slope, coc = coconut mat)

Maximum mean pressure [mmHg]											
FOAM						JAY ACTIVE™					
lin	pvmt	low	curb	high	coc	lin	pvmt	low	curb	high	coc
143	133	133	122	116	100	70	77	76	75	72	-
178	177	210	207	213	209	101	100	96	101	96	101
164	167	191	196	176	164	94	100	97	116	86	80
199	150	265	248	170	157						
146	141	138	135	134	106						
246	234	240	232	249	134						

A two-way analysis of variance on the maximum mean pressure produced the following results (Table 6-3).

Table 6-3. Two-way analysis of variance of the maximum mean pressure under the buttocks of nine wheelchair users, using two types of cushions on six different conditions

Source of variation	Sum of Squares	d.f.	Mean Square	F-ratio	Sig. Level
Main effects					
A(cushion)	94052.17	1	94052.17	53.94	<0.001 S
B(condition)	4227.80	5	845.56	0.476	0.79 N.S.
Interactions					
AB	2567.22	5	513.44	0.289	0.92 N.S.
Error	72837.50	41	1776.52		
Total	172199.17	52			

The results showed no significant interaction between cushions and conditions. The maximum mean pressure showed no significant variation due to the six different conditions (notice that they were driven in the same order each time). The two cushions, a foam cushion and a Jay Active™ cushion, showed a highly significant difference ($P<0.001$), with the lowest pressure acting on the Jay Active™.
Subsequently, the ratio of the pressure below (oxygen recovery R) and above (oxygen debt D) the closing pressure of 100 mmHg (13.3 kPa) was analysed. For the different measurements in the different conditions the following was found.

Table 6-4. Number of observations of tissue response according to the R/D ratio as defined by Bar[1]. In all cases a foam cushion produces severe tissue response

	Cushion	
Response	Foam	Jay Active™
Mild	0	12
Moderate	0	4
Severe	36	1

As can be seen in Table 6-4, the foam cushion produced a severe response of the tissue under the buttocks in all cases. In most cases Jay Active™ produces a mild response.

Discussion

The measurements showed that it is correct to assume that wheelchair driving is a dynamic activity, leading to continuous pressure variations under the buttocks. Static measurements would have produced one pressure varying from 0 to 200 mmHg (0-26.3 kPa), depending on the posture of the subject during the measurement.
Static measurements on wheelchair cushions produce often higher pressures than the maximal capillary arterial blood pressure reported by Landis, while the cushion is well known as a prevention for pressure

sores in the clinical practice[11-14]. To consider wheelchair use as a dynamic activity, leading to continuous pressure variations, could be an explanation for this contradiction.

A special wheelchair was used with a mass of 33 kg. Driving this wheelchair was heavier than a standard wheelchair, therefore the pressure might be higher during this test.

The conditions of the test circuit were driven in the same order by every subject, because of other measurements that were performed simultaneously with our pressure measurements. This makes comparison between each condition doubtful. But all conditions together were set up to be the mean wheelchair users situation, and the total test gives an insight herein.

In the rehabilitation center we co-operated with, for some patients a foam cushion could be used as a pressure relief cushion. The measurements do not support these findings, but a less severe R/D quotient might be found for these subjects when the measurements will be performed during a period longer than 3 minutes (2 hours, for example).

The Jay Active™ was prescribed in the majority of cases as a cushion to prevent pressure sores. These findings are supported by the measurements.

Conclusions

Dynamic pressure measurements were performed on nine subjects with spinal cord injuries. The data were processed using the findings of Bar[1]. Bar calculated the quotient of pressure-time measured below and above a threshold pressure of 100 mmHg (13.3 kPa). This quotient correlates with three classes of tissue response under the buttocks: severe, moderate and mild.

The data showed that a 5 cm thick foam cushion produced severe tissue responses in all cases, while a Jay Active™ cushion showed mild responses most of the time.

The maximum mean pressure showed no significant variation due to the six different driving road-conditions (it should be noted that they were driven in the same order each time).

The advantage of this method is that dynamic measurements give better information than static ones. A static measurement gives only one pressure distribution, based on one posture of the wheelchair user, while the effect on pressure caused, for instance, by driving is neglected. In this study the variations of posture while driving a wheelchair on different surfaces were included in the data; therefore this method seems promising as an aid to study the effect of pressure relief cushions.

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Chapter 7

The curvature of the lumbar spine in sitting

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Submitted

The curvature of the lumbar spine in sitting

Summary

One of the purposes of a good chair is to prevent continuous overload of the lumbar spine. In the study described here the lower lumbar curvature was measured during a 55 minute reading task when seated in an office chair with a forward and backward tilted seat, and on a "balans" chair.
On all these chairs the lumbar curvature becomes more kyphotic (flattens) compared to standing.
The mean lumbar curvature during a 55 minute reading task is significantly more lordotic on the "balans" compared to an office chair with a forward tilted seat.
There is no significant difference in mean lumbar curvature during a 55 minute reading task when sitting on a "balans" chair or an office chair with a backward tilted seat.

Introduction

It is the purpose of a good chair to prevent continuous overload of the lumbar spine and its resulting back problems. Different studies have shown that the lumbar spine is flattened when a subject sits in a conventional office chair. The 90° flexion between trunk and thighs is established through 60° flexion of the hips while the remaining 30° comes from flattening of the spine[1-3]. Keegan[2] studied the lower lumbar curvature by means of röntgenograms and concluded that a normal curvature is adopted when the angle between trunk and thighs equals 135°. When sitting with an angle between trunk and thighs of 90°, extrusion of the lower lumbar disks occurs, leading to substantial back problems. From his study Keegan concluded that a chair should at least have an angle between backrest and seat of 105°. To establish contact with the backrest he advised a seat angle of 5°.
Mandal[4], however, concluded from his own studies and the observations of Åkerblom, Keegan and Schoberth that a seat should not be tilted backwards, but on the contrary be tilted forward. He advised a forward

tilted seat at an angle of 20°. He found that only then the lower lumbar spine of the subject is less flattened and the back problems might be reduced. The chair also has knee rests attached to it and is known as the "balans" chair.

Since the introduction of the "balans" chair several studies have been carried out using this seat. Brunswic[5] monitored subjects that were asked to sit upright on a "balans" chair and a conventional chair by means of a hydrogoniometer. She concludes that a decreased hip flexion adopted in the "balans" chair provokes less lumbar strain.

Frey and Tecklin[6] have also compared the lower lumbar curvature on a "balans" chair, a conventional chair and during standing. They measured the curvature by means of a flexible ruler immediately after the subject had claimed to sit comfortably. They found that there was less flattening of the lower lumbar curvature when sitting on a "balans" chair.

Bendix[7] used the statometric method to evaluate work station variables and found that increased seat height and inclination forward increased lumbar lordosis, decreased kyphosis. These studies confirmed the advantage for the lower lumbar curvature of the "balans" chair. The main disadvantage of these studies, however, is that they only measure the curvature at one point in time, while it can be expected that during sitting the posture and thus the curvature change. This was confirmed by the study of Lander *et al.*[8]. They measured the EMG on the cervical and paraspinal surface on a conventional office chair and a "balans" chair and found a gradual increase in myoelectric amplitude in subjects sitting in a "balans" chair. They stated that this is caused by the absence of a backrest.

Due to the fact that the studies so far of the lower lumbar spine were static and because an alteration of the lumbar spine can be expected in time, in this study the lower lumbar curvature is monitored continuously during sitting on a "balans" chair and a conventional office chair. In order to examine if the claim of less flattening of the curvature on a "balans" chair holds for longer periods of time, a special measuring device was used. Furthermore, a biomechanical model is used to gain insight into the posture and the forces that act on the trunk during sitting.

Biomechanical model

Before studying the aspects of seating with the aid of a biomechanical model, the sitting posture is compared to standing. Measurements[1-3] show that the lumbar spine in sitting is always flattened compared to standing. When standing, the pelvis is supported by the head of both femurs and equilibrium of moments about the hip joint is provided for by leg muscles. In sitting however, the pelvis is supported by the tuberosities and it may be expected that the psoas major muscle shows less or no activity. As a result, the lumbar curvature flattens in sitting. Other explanations of lumbar flattening[3] refer to pelvic tilt as a result of tension in the ischiocrural muscles when sitting.
To gain insight in the forces that act on the trunk during sitting a biomechanical model is presented. A cross section of the trunk is made at the level of the discus L5. The mass centre of gravity of the part of the body above this level is located near the arm pits. The model is restricted to the sagittal plane. The influence on the disc load and posture is evaluated in four sitting situations; one with active back muscles and three without back muscle activity.

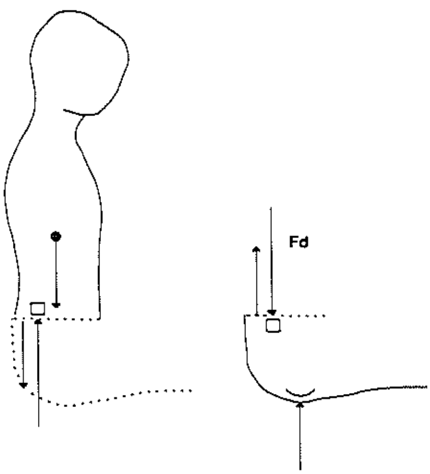

Figure 7-1. Equilibrium of forces in an active unsupported upright position.

When sitting upright in an unsupported posture, the centre of gravity of the trunk is situated ventrally to L5. In order to retain the upright position, the back muscles must be active, see Figure 7-1.

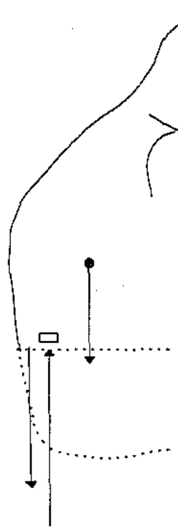

Figure 7-2. Equilibrium of forces in an upright unsupported slumped posture. The ligaments establish the equilibrium and the disc load is higher than in the active upright position.

In the posture described above, the back muscles must be active continuously. In the passive upright unsupported posture, ligaments are used to relax the muscles. The tension on the ligaments is established by tilting the pelvis backwards and by flexion of the spine, Figure 7-2. This increases the distance from the centre of gravity towards the intervertebral disc (L5) and the moment of force with respect to the disc is larger than in active upright sitting. To establish equilibrium, the ligament forces must therefore be higher than the muscle forces. This results in a higher disc load, see Figure 7-2. In the above mentioned posture it is assumed that the resultant ligament force has an equal lever arm in relation to the intervertebral disc, compared to the muscle force.

This lever arm will be longer with a considerable contribution of the thoracolumbar fascia. The precise force distribution, however, remains obscure.
The arms can also contribute to the equilibrium of the trunk to remain in an upright position without using the back muscles, see Figure 7-3. Because the arms support a part of the trunk, the disc load is lower than sitting in the unsupported upright position. When it is assumed that the pelvis is supported by the seat at the ischial tuberosities, the reaction force of the spine on the pelvis induces a couple of force that rotates the pelvis backwards. To prevent the pelvis from tilting, a force on the crista iliaca is desired. This force can be provided for by a backrest. When the backrest is not properly used, the pelvis will turn backwards until the tension in the ligaments establishes equilibrium. To prevent the pelvis from tilting it is also possible to bend the trunk forward until F_d is situated above the ischial tuberosities.

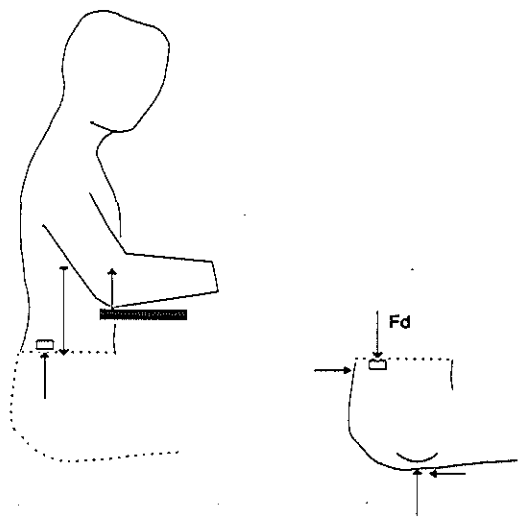

Figure 7-3. Equilibrium in supported upright sitting. The force on the arms lowers the load on the disc.

Another possibility to sit upright, without activity of the back muscles is to tilt the trunk until the centre of gravity is above the disc L5, see Figure

7-4. This posture, however, requires activity of leg muscles. Therefore a supporting force at the crista iliaca is needed, both to keep this posture and to prevent the pelvis from tilting backwards.

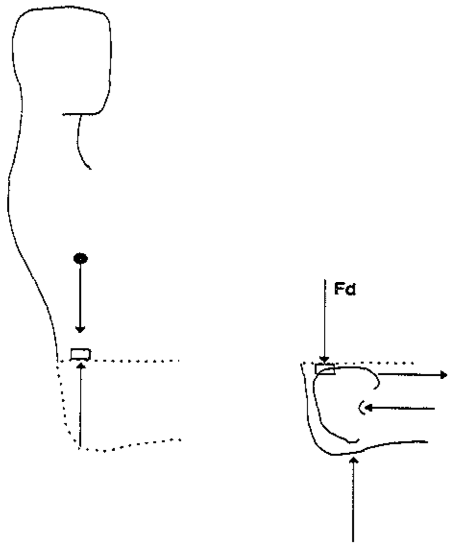

Figure 7-4. Equilibrium in supported upright sitting. The centre of gravity of the trunk is situated above the disc L5, but far behind the ischial tuberosities.

By considering the model the following can be concluded. When sitting upright without any support, the back muscles must be active continuously. When no backrest is available, the muscles can be relaxed by depending on the ligaments for postural support. Because the ligaments must be tensed by tilting the hip and bending the spine, this will lead to a kyphotic low back.

When the arms are used to support the trunk, the reaction force of the spine on the pelvis induces a couple that rotates the pelvis backwards, leading to a kyphotic low back, as well. Only a properly used backrest, that gives support on the crista iliaca, prevents pelvic tilt.

Therefore, the authors believe that a chair without a backrest ("balans" chair) will always lead to a more kyphotic posture than during standing. A chair with a backrest, however, when properly used can provide for a diminished kyphotic low back.

Materials and methods

In this study, the curvature of the low back is defined as the angle ϕ between two lines perpendicular to the back through the disci S2 and Th12, see Figure 7-5.

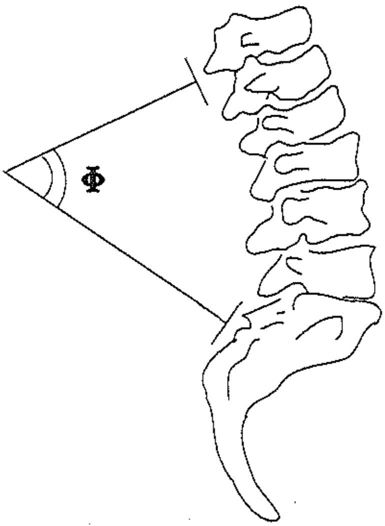

Figure 7-5. Definition of the angle ϕ between two lines perpendicular to the back through S2 and Th12.

A special device was used to measure this angle[9] which will be briefly described. Part of this device consists of two springs of which the length is measured by means of strain gauge type force transducers. When the ends of the springs are positioned on the back between S2 and Th12, it was shown by previous measurements that there is a linear relationship between the length of the springs and ϕ, Figure 7-6.
It was also shown that a backrest could be used comfortably with the springs taped on the back of the subjects. In order to measure ϕ, two inclinometers were taped on Th12 and S2. The inclinometers measure the angle to the horizontal and ϕ can be calculated by means of the following formula:

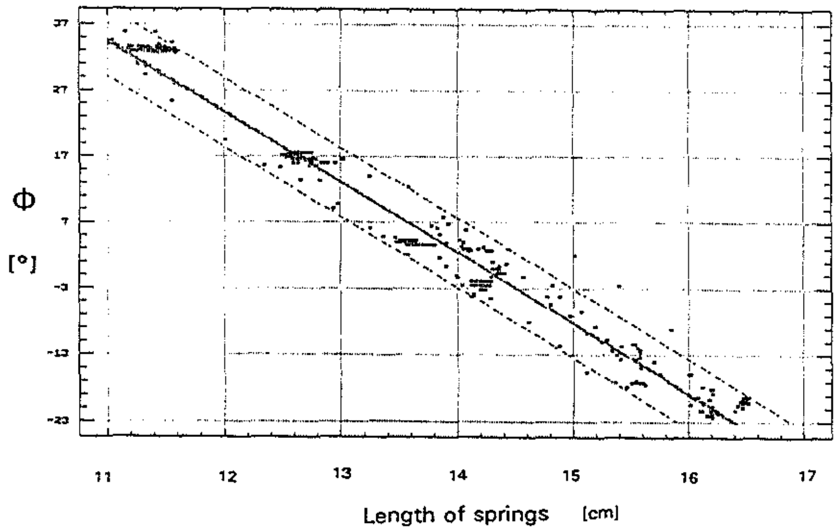

Figure 7-6. The linear relationship between the length of the springs and ϕ.

ϕ = angle upper inclinometer- angle lower inclinometer + offset

A negative value of ϕ means that the lower lumbar curvature is kyphotic, and a positive ϕ means a lordotic lower lumbar curvature. The offset was calculated so that when the two inclinometers were held under the same angle, ϕ equals 0.

With the springs and the inclinometers, the length and the curvature (ϕ) of the lower lumbar back could be measured at the same time, Figure 7-7.

To calibrate the different curvatures, the subjects were asked to bend forward from their hips, the height of the hands above the ground were measured at several increments going forward. This was repeated for the subject returning to a standing position from a forward bent posture.

The linear relationship between ϕ and the average length of both springs was determined with the software package Statgraphics 5.0. After this the inclinometers were removed, and the subjects could use the backrest of the seat without hinder from the springs as they are only 5 mm thick.

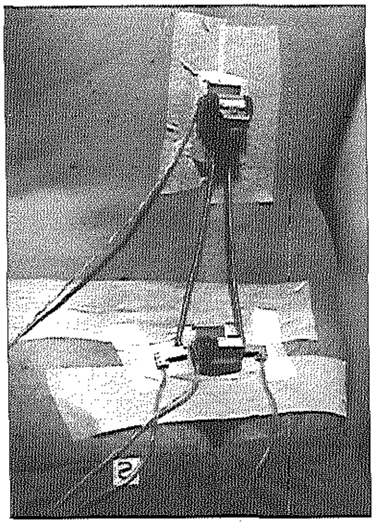

Figure 7-7. Attachment of the springs and inclinometers on the back of a subject.

Two hypotheses were tested:

1. Compared to the curvature during standing (positive value), the lumbar curvature flattens (becomes more negative) on as well an office chair as a "balans" chair.

   H0: $\Phi_{\text{"balans" chair}}, \Phi_{\text{forward}}, \Phi_{\text{backward}} = \Phi_{\text{standing}}$

   H1: $\Phi_{\text{"balans" chair}}, \Phi_{\text{forward}}, \Phi_{\text{backward}} \leq \Phi_{\text{standing}}$

2. When sitting on a "balans" chair and sitting on an office chair with the seat tilted forward or backwards, the average low back curvature of the "balans" chair is not equal compared to the other two situations.

   H0: $\Phi_{\text{"balans" chair}} = \Phi_{\text{forward}} = \Phi_{\text{backward}}$

   H1: $\Phi_{\text{"balans" chair}} \neq \Phi_{\text{forward}}, \Phi_{\text{backward}}$

To test the hypotheses, eight healthy subjects without a history of back complaints and of which 5 female, were measured (age 22, SD= 1.1 year, height 1.76, SD= 0.07 m). Two types of chairs were used, namely

the "balans" chair (VARIABLE balans, STOKKE) and an office chair (CREDO 2260, HAG). The seat of the office chair could be fixed under an angle of 5° forward and backwards. An office desk was used of 76 cm height with a 10° inclined reading surface[10] on which a book was placed. The subject had to read, while sitting on each chair for a period of 55 minutes, during which they were not allowed to remove the book from the desk. They were sitting in a quiet room and could not be disturbed by telephone or otherwise. After this period the subject had a break for 5 minutes. The chairs were used in random order.
To collect the data from the springs, the Ramcorder was used. The Ramcorder is a device that is developed within our department and that is capable of storing processed data from 8 input channels into digital memory. The signals were sampled with a frequency of 8 Hz. To calculate the average ϕ over the period of 55 minutes, the software package CODAS, version 5.50 was used. The hypotheses were tested comparing the mean curvatures in pairs in Statgraphics 5.0, using a paired t-test with a level of significance (α) of 0.05.

Results

In Figure 7-8 a typical example of a continuous recording of the lumbar curvature is given. The average of low back curvatures ϕ in the four situations are presented in Table 7-1.

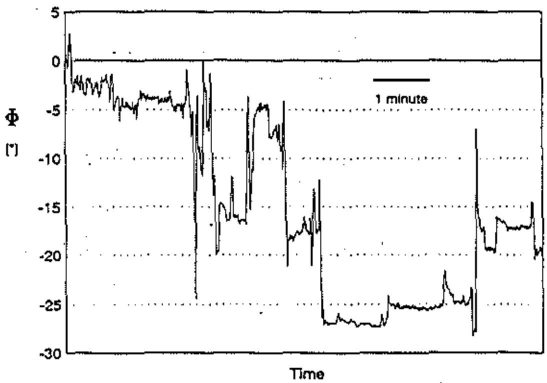

Figure 7-8. A typical example of a continuous registration of the lumbar curvature.

Table 7-1. Average lower lumbar curvature of eight subjects during a 55 minute reading task

Subject	Standing [°]	Knee [°]	Forward [°]	Backward [°]
1	27.9	14.3	5.5	6.2
2	19.2	-8.9	-6.8	-5.1
3	24.6	-0.1	-8.9	-3.4
4	27.4	-3.1	-18.7	-11.4
5	26.7	3.8	8.6	9.6
6	29.6	1.9	-1.2	-2.9
7	29.1	10.4	-7.3	-3.1
8	33.4	18.8	13.1	8.3
mean	27.2	4.6	-2.0	-0.2

It can be seen from the data that, without exception, the low back curvature ϕ decreases (becomes more kyphotic), when sitting. With the paired t-tests the following was found.

Table 7-2. Comparison of low back curvatures ϕ in two different situations. Both means are compared by means of a paired t-test. The P-value denotes whether the difference is significant (S)

ϕ_{mean}	ϕ_{mean}	P
standing	"balans" chair	< 0.0001 (S)
standing	forward	< 0.0001 (S)
standing	backward	< 0.0001 (S)
"balans" chair	forward	0.05 (S)
"balans" chair	backward	0.08 (N.S.)
forward	backward	0.24 (N.S.)

The first hypothesis, stating that the low back curvature in sitting is the same compared to standing, is rejected by the data. The low back curvature is more kyphotic in all situations (alternative hypothesis).
The second hypothesis, stating that the low back curvature is not significantly different on the three chairs, is also rejected by the data. It is shown that the curvature is more lordotic, when sitting on the "balans" chair compared to a forward tilted office chair. It is also shown that the curvature is not significantly different, when sitting on a "balans" chair compared to a backward tilted office chair.

Discussion

The form of the lumbar spine is often taken as the most important biomechanical aspect in sitting. Prevention of pronounced lumbar kyphosis by means of a forward tilt of the seat was advocated by Mandal[4], and in one of the chair designs a knee support was introduced to prevent the subject from sliding forward. The claim of lumbar lordosis was supported by static measurements[6]. In addition to these studies we measured in dynamic situations for a prolonged period of time. For this purpose we placed sensors on the lower back, which did not hinder the subject, nor disturbed the use of the backrest. Up to now these sensors were only used above the lumbar level.
The data illustrates that every subject has a different low back curvature when sitting. For example, some of the subjects have a kyphotic low back on the "balans" chair, others a lordotic one. This is consistent with other findings in literature. Frey and Tecklin[6] studied 44 healthy subjects on a "balans" chair and a conventional office chair. They found a range of low back curvatures varying from -36° to 9° on a conventional office chair, and from -27° to 32° on a "balans" chair.
The data of the present study show that there is a continuous change of curvature on the "balans" chair as well as the normal office chair (Figure 7-8). This observation raises the question whether one measurement in time is sufficiently accurate to study the lower lumbar cuvature when sitting.
The measurements support the findings of other authors that there is a flattening of low back curvature when sitting is compared to standing.

The availability of a backrest, however, does not lead automatically to a more lordotic posture. On the contrary, the "balans" chair seems to give the most lordotic posture during the reading task. This can be explained by not using or incorrect use of the backrest by the subjects during the reading task.
In the present study, only the curvature of the lower lumbar back was studied. Muscle activity of the back muscles and the increased load of the knee on the "balans" chair was not studied. Lander[8] found increased cervical and lumbar muscle EMG after sitting in the "balans" chair compared to a conventional office chair.
As reading tasks form a limited part of many sedentary jobs, the influence of different seat inclinations should be studied in realistic situations during whole working days. The study of Lander[8] shows that other conclusions may be expected due to other tasks and fatigue.

Conclusion

Measurement of the lower lumbar curvature during a 55 minute reading task showed that the mean curvature is more kyphotic compared to standing when sitting on an office chair with a forward and a backward tilted seat, and on a "balans" chair.
The mean lumbar curvature during a 55 minute reading task is significantly more lordotic on the "balans" compared to an office chair with a forward tilted seat.
There is no significant difference in mean lumbar curvature during a 55 minute reading task when sitting on a "balans" chair or an office chair with a backward tilted seat.

Acknowledgements

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Appendix A

Dimensions for office chairs

Dimensions for office chairs

In this appendix a review of the literature is presented on the preferred dimensions for office chairs. Although this list is not complete, it represents the well known sources published up till now. Wherever possible, the motives underlying the dimensions are given.
Moreover advice is given for each dimension based on the assumption that the office chair should be suitable for P5 to P95 Dutch adults as presented in the Dined-table.
All dimensions are defined according to NEN 1812, *Ergonomic requirements for office work-chairs and office workdesks, Requirements for dimensions and design, Measurement and test methods.* Figure A-1 shows the definition of the dimensions.

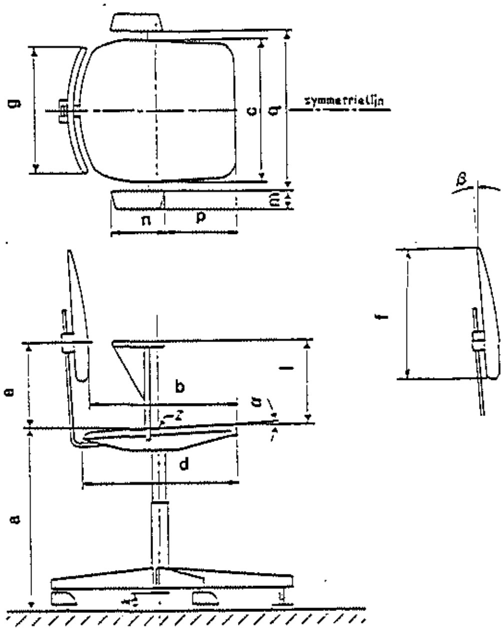

Figure A-1. Definition of the dimensions for office chairs. (source: NEN 1812)

Seat height (a)

Author	Range	Reason
Åkerblom, 1948 op.cit.in Burandt, 1963	38-41 cm	Discomfort of thighs is expected above 41 cm.
Floyd, 1959 op.cit.in Burandt, 1963	43-46 cm	?
Keegan, 1962	41 cm	Discomfort of thighs is expected above 41 cm.
Burandt, 1963	45-53 cm, at table height 78 cm 40-48 cm, at table height 72 cm	Preferred by subjects.
Grandjean, 1973	43 cm	Preferred by subjects.
Maybey, 1977 op.cit.in Bendix, 1987	below popliteal level	Preferred by subjects.
Mandal, 1982 op.cit.in Bendix, 1987	10-15cm above popliteal level	Preferred by subjects.
Bendix, 1987	3-5 cm above popliteal level	More lordotic lower lumbar curvature. Preferred by subjects above low and backward tilted seat.
Goetschel, 1987	41-57 cm	Both feet on the floor.
Sanders, 1987	43 female, 46 male	P5 knee height increased with shoe height of 2.5 cm.
Chi-Yuang, 1988	51-61 cm	Preferred by subjects.
Oborne, 1988	43-50 cm	Both feet on the floor. Low pressure under thighs.
NEN 1812, 1990	39-51 cm minimal range	knee height

The seat height according to NEN 1812 is based on the centre of the seat. Most authors, however, base this height on the front of the seat. For an office chair with a backward tilted seat, as a rule of thumb 2 cm can be added to the NEN-value to obtain the height to the front of the seat.

Bendix considers the height and inclination of the seat covariates. Out of the literature presented above it can be seen that a forward tilted seat requires a seat height above popliteal level and a backward tilted seat a height at popliteal level.

The disadvantage from a forward tilted seat is a shear force that acts on the buttocks of the sitter, increased muscular activity and the absence of a backrest. The advantage is claimed to be a decreased kyphotic lower lumbar curvature. In this thesis, however, during a 55 minute reading task no significant difference in mean lumbar curvature was found when sitting on a "balans" chair or on an office chair with a backward tilted seat.

Based on the current knowledge, a backward tilted seat for office chairs is preferred. The front of the seat must be so that it is possible to have both feet flat on the ground. When a shoe height of 3 cm is added, the seat height (measured at the front of the seat) should be:

seat height: 39-52 cm

Seat depth (b)

Author	Range	Reason
Burandt, 1963	35-40 cm	Preferred by subjects.
Shackel, 1969	38-42 cm	Preferred by subjects.
Grandjean, 1973	<43 cm	P5
Croney op.cit.in Panero, 1979	34-38 cm	?
Diffrient op.cit.in Panero, 1979	38-41 cm	?
Bendix, 1987	2/3 of thighs	Not too much restriction in freedom of movement.
Goetschel, 1987	36-47 cm	Support thighs. Not too much restriction in freedom of movement.
Oborne, 1987	35-40 cm	P5
NEN1812, 1990	38-44 cm minimum range, maximum 47 cm	?

To support the thighs as much as possible, as a rule of thumb four fingers must fit between back of the knee and front of the seat. Another aim is to enable contact with lumbar support. When 4 cm is assumed, the seat depth must be:

seat depth: 41-52 cm

Effective seat depth is meant here. Raised brims on the seat or too large roundings are not used for support and must therefore not be added to the seat depth.

Seat width (c)

Author	Range	Reason
Croney op.cit.in Panero, 1979	43 cm	?
Diffrient op.cit.in Panero, 1979	41 cm	?
Dreyfuss op.cit.in Panero, 1979	38 cm	?
Panero, 1979	43-48 cm	?
Oborne, 1987	40-43 cm	P95, female.
Sanders, 1987	>40 cm	Preferred by subjects.
NEN 1812, 1990	40 cm, minimum	?

The seat must support the entire buttocks. Therefore it must be dimensioned on the P95, and must be:

seat width: 43 cm, minimum

Seat inclination (α)

Author	Range	Reason
Keegan, 1953	5°, backward	To encourage use of the backrest.
Burandt, 1963	3°, backward	Preferred by subjects.
Mandal, 1974	30°, forward	To establish a lordotic lumbar curvature.
Bendix, 1987	forward, free adjustable	Preferred by subjects.
Goetschel, 1987	5-15°, forward 5°, backward	To establish a lordotic lumbar curvature.
Bridger, 1988	25°, forward	Preferred by subjects above conventional office chair after two weeks.
Chi-Yuang, 1988	forward	Preferred by subjects.
Oborne, 1988	<3°, backward	To encourage the use of the backrest.
NEN 1812, 1990	2-4°, backward (not adjustable) 17° backward- 4° forward (adj-ustable)	To encourage the use of the backrest.

In this thesis it was found that, when little shear is accepted, the angle between seat and backrest must be between 90 and 95°.

seat inclination: **so that angle between seat and backrest equals 90-95°**

Height of the lumbar support (e)

Author	Range	Reason
Burandt, 1963	19-23 cm	Preferred by subjects.
Croney op.cit.in Panero, 1979	13-19 cm	?
Diffrient op.cit.in Panero, 1979	23-25 cm	?
Dreyfuss op.cit.in Panero, 1979	18-28 cm	?
Panero, 1979	19-25 cm	?
Chi-Yuang, 1988	25 cm	Preferred by subjects.
NEN 1812, 1990	17-21 cm minimum range	Experiments ?

The lumbar support diminish a kyphotic form of the lower spine. To be able to do so, it is necessary to know the height of the maximum concavity of the lumbar curvature above the ischial tuberosities. However, very few papers exist on this subject. Darcus found for British males a range from 20-30 cm, but used 27 cm above the seat in practice. Diffrient states 23-25 cm. Diebschlag found in a study on 91 subjects that the height of the crista for P5-P95 was 18-25 cm. Therefore 18-25 cm seems to cover most of the ranges that are meant in the papers known to us.

Height of the lumbar support: 18-25 cm

Height of backrest (f)

Author	Range	Reason
British Standard op.cit.in Frey, 1986	<33 cm	?
Croney op.cit.in Panero, 1979	10-20 cm	?
Diffrient op.cit.in Panero, 1979	15-22 cm	?
Grandjean op.cit.in Panero, 1979	20-30 cm	?
Panero, 1979	15-23 cm	?
Goetschel, 1987	>50 cm	Preferred by subjects.
Chi-Yuang, 1988	>15 cm	Preferred by subjects.
Oborne, 1988	48-63 cm	?
NEN 1812, 1990	22 cm maximum, low 37 cm minimum, high	?

The backrest is one of the most difficult aspects of a chair. Branton concluded from a study on 114 seated persons that it was disappointing to find that ranges in the vertical are so great. As an example he stated that a head rest should have to be adjustable over a wide range and then only about 20-30% of the sitters were satisfied. Moreover, the horizontal variation of the thorax is 4-6 cm and too large to be compensated for by any softness or cushioning. From the literature it can be seen that a low backrest varies from 15 to 30 cm and a high backrest is up to the shoulders, 48-63 cm.

Most authors agree that there should be free space for the buttocks between the seat and the backrest (Snijders, > 12 cm).

The high backrest only provides support in relaxed postures (backrest angle > 20°) and in upright sitting postures it restricts the movements of the sitter and is believed to force the sitter in a kyphotic posture. Therefore, in office chairs only a low backrest is advisable:

Height of the backrest: 15-30 cm (free space seat-backrest: > 12 cm)

Width of the backrest (g)

The width of the backrest is often not explicitly mentioned, but it can be concluded that the backrest is as wide as the seat or slightly smaller.

Inclination of the backrest (ß)

Author	Range	Reason
Keegan, 1953	20-40°	To establish a lower lumbar curvature.
Andersson, 1974	>20°	Low intradiscal pressure and low muscle activity.
Croney op.cit.in Panero, 1979	5-25°	?
Diffrient op.cit.in Panero, 1979	5°	?
Dreyfuss op.cit.in Panero, 1979	5-15°	?
Panero, 1979	5-15°	?
Goetschel, 1987	20-30°	Low intradiscal pressure and low muscle activity
Grandjean op.cit.in Oborne, 1987	11-14° (reading) 15-18° (relax)	Comfort
Oborne, 1987	13-22°	Highest comfort at 18°.
Sanders, 1987	10-30°	?
Zacharkow, 1988	5-10°	Optimal allert sitting posture.

From literature it can be seen that for an upright active position the backrest needs to be tilted 5-15°. Based on the assumption that low discal pressure and low myoelectric activity is preferable, Andersson found that a chair should have an angle between seat and backrest of at least 100°. When the seat has an inclination of 5°, this would mean a backrest inclination of 15°.

Backrest inclination: 5-15°

Height of the armrest (l)

Author	Range	Reason
Panero, 1979	21-23 cm	?
AT&T op.cit.in Sanders, 1987	25 cm	?
Goetschel, 1987	20-25 cm	Support during sitting down and standing op.
Oborne, 1987	20-24 cm impressed seat	Support when changing position.
NEN 1812, 1990	20-27 cm adjustable 23-25 cm non-adjustable	P5-P95

The armrests have several functions. They relieve the muscles in the shoulders and neck, support the upper body and reduce the pressure on the seat. Moreover, they give support when sitting down and standing up and during a change in position.
According to the Dined-table, P5-P95 seat-elbow height is 20-28 cm. To support also during anteflexion an additional 5 cm can be added:

Height of the armrest: 20-33 cm

Width of the armrest (m)

Not many studies report about the width of the armrest. Mostly 5 cm is mentioned as a minimum, based on the width of the forearm.

Width of the armrest: 5 cm, minimum

Length of the armrest (p)

Although most authors mention armrests as important aspects of an office chair, the dimensions are seldom mentioned.

Author	Range	Reason
Panero, 1979	25-31 cm	Measured from backrest.
Goetschel, 1987	25 cm	Measured from backrest.
NEN 1812, 1990	20-24 cm (adjustable) 10 cm minimum (non-adjustable)	To support the elbow.

During writing the armrests must give support to the elbows, and it must also be possible to draw up the seat to the table. P5-P95 body depth, calculated from DIN 33402 according to the method used in the Dined-table, gives 22-34 cm. When 5 cm is added as extra space, the distance from backrest to front of the armrests should be:

Length of the armrest: 27-39 cm

Width between armrests (q)

The width between the armrests must be so that there is not too much abduction during sitting and also no hinder during sitting down and standing up. P5-P95 elbow width is 39-54 cm. P95 width of the pelvis with an additional 3 cm, is 47 cm.

Width between armrests: 54 cm, non adjustable
39-54 cm, adjustable

Summary

In the following an overview can be found of the advised dimensions.

Table A-1. Advised dimensions for office chairs

Chair dimension	Range
Seat height	39-52 cm
Seat depth	41-52 cm
Seat width	43 cm, minimum
Seat inclination	so that angle seat-backrest equals 90-95°
Height of the lumbar support	18-25 cm
Height of the backrest	15-30 cm (free space seat-backrest > 12 cm)
Width of the backrest	seat width
Backrest inclination	5-15°
Height of the armrest	20-33 cm
Width of the armrest	5 cm, minimum
Length of the armrest	27-39 cm
Width between the armrests	54 cm, non-adjustable 39-54 cm, adjustable

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Appendix B

Products and Prototypes

Products and Prototypes

During the study presented in this thesis, various projects mostly in cooperation with companies were carried out, each resulting in a new product, prototype or an improved product. In the following a brief overview of products is presented. Not included in this appendix are the evaluation of several chairs and cushions for Dutch and German companies.

Figure B-1. A low-cost anti-decubitus mattress for home care. A prototype was developed for Linido in cooperation with Ejok Design for Industry. Patent is pending for this prototype.

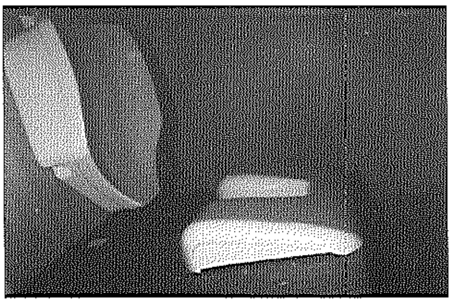

Figure B-2. Adaptations were implemented in a seat to be used under a shower. Based on measurement of the supporting forces on seated subjects. The seat is manufactured by Linido.

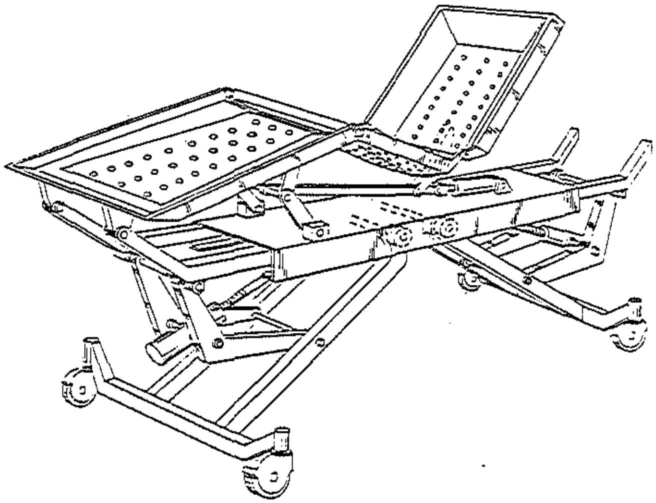

Figure B-3. Based on the findings presented in this thesis a prescription was given for the seat- and backrest inclinations of a hospital bcd. Schell implemented these findings in their new hospital bed, which is currently produced in large series.

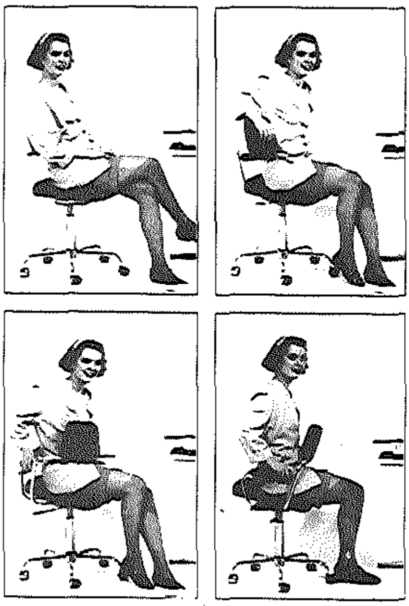

Figure B-4. With the aid of the measuring instrument introduced in chapter 2 the angles for a dentist chair were prescribed and implemented by Ergo Design. The dentist chair is on the market.

Figure B-5. Advice was given for the design of chairs to be placed in lecture-halls. A. Dorscheidt developed a chair partly based on this knowledge. The first series will be placed in the new M-building of the Erasmus University of Rotterdam in the middle of 1994 and is manufactured by Eromes.

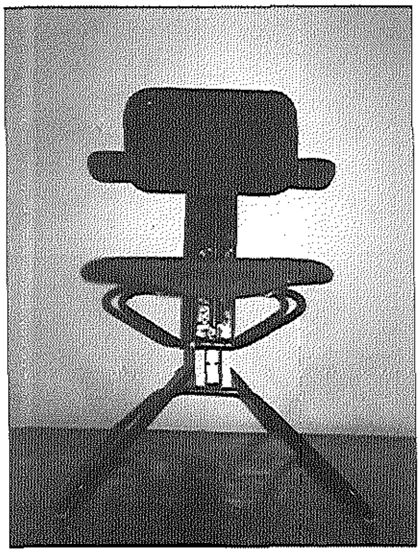

Figure B-6. Advice was given for the design of a chair for musicians. A prototype was developed by N. van Haaster for Grahl. The chair was presented at Interoffice '93.

Figure 8-7. Advice was given for the development of a chair for microsurgeons. This complex chair was designed by J.W. van Rhijn for BMA Ergonomics. The prototype is in daily use at Dijkzigt Hospital.

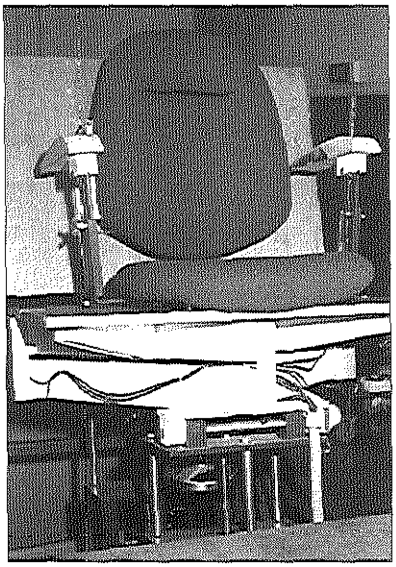

Figure 8-8. A first concept of a new office chair from Ahrend was evaluated and suggestions were made for improvement.

General discussion and summary

General discussion and summary

In modern western society more and more time is being spent sitting on a chair during working hours as well as leisure time. One of the reasons that special attention is given to this field is the enormous number of back complaints. Approximately 80% of American adults suffer from some disorder of the lumbar spine during their lifetime. It is assumed that design of the furniture is an important factor in the aetiology. Although the issue seems simple, even the choice of furniture, at home as well as in professional use, raises difficulties. We attribute this to a lack of existing knowledge of design criteria of furniture. Moreover the available knowledge often is under discussion and based on scarce data.

We observe also that in many aspects research is either lacking or insufficiently thorough, due to the absence of apparatus to measure complex quantities in vivo. With this in mind various new pieces of laboratory apparatus and a portable posture registration set (PPRS) have been developed at the Department of Biomedical Physics and Technology of the Erasmus University of Rotterdam. These computer integrated apparatus allow for the acquisition of new data on complex physical quantities. In this thesis the equipment is used to focus on load distribution, local pressure and shear and the form of the spine in relation to individual characteristics, form and dimensions of chair design and material properties of body supporting surfaces.

As such, this study concerns a number of projects, each providing new insight in the consequence of body support by testing and quantifying hypotheses often discussed in literature. The projects are presented in different chapters which are summarized as follows.

Knowledge of the interaction of forces between a person and the bed in which they lay or the seat on which they sit, provides an insight into the loading of their muscles, bones and soft tissue.

In **chapter 2** a new instrument is introduced, by which the forces perpendicular and parallel to three different freely adjustable body supporting surfaces (backrest, seat pan, foot support) can be registered.

Forces were measured on a configuration of these surfaces simulating a bed with the backrest at an angle of 45° to the horizontal and the

mattress horizontal. The measurements on healthy subjects (mean mass 77, SD= 11 kg) showed an accuracy of ± 10 N. In this position the mean shear force on the seat pan was 97 N.
After quantification of the shear force on the buttocks in the "seated in bed" posture, the next study focuses on the local distribution of the shear force on the skin.
Shear is understood to be one of the important factors in the development of pressure sores. In **chapter 3** a new sensor is introduced that can measure between skin and deformable materials without disturbing the shear phenomenon. Validation experiments were performed on a series of ten healthy young subjects in which it was demonstrated that the sum of the local shear forces is equal to the resultant shear force measured with a force plate. When sitting in a foldable wheelchair shear stress of 4.8 kPa was recorded in the longitudinal direction and 8.5 kPa in the transversal direction. On a mattress of a hospital bed shear stress was found of 5.6 kPa in the longitudinal direction and 3.1 kPa in the transversal direction.
From literature it was known that, when shear stress was at a sufficiently high level (9.9 kPa), the combination of pressure and shear was found particularly effective in promoting blood flow occlusion in the palm of the hand. This raised the question: What influence does the lowest shear stress that was measured with the new sensor (3.1 kPa), have on the blood flow?
In **chapter 4** skin oxygen tension is presented as a measure for blood flow. This tension was measured at the sacrum with application of combined pressure and shear stress on the skin of young healthy subjects (mean age 25.5, SD= 3.4 years).
Cut-off pressure is defined as the level of external pressure exerted on the skin at which the skin oxygen tension is 1.3 kPa; at this level ischemia of the skin can be expected. A mean cut-off pressure of 11.6 kPa was found when no shear stress is applied versus a significant lower mean cut-off pressure of 8.7 kPa with a shear stress of 3.1 kPa. No significant relationship was found between cut-off pressure and systolic blood pressure, diastolic blood pressure, skin thickness at the sacrum, percentage of fat and skin oxygen tension in the unloaded situation.
From the above it was concluded that the inclination of backrest and seat are important design criteria. In **chapter 5** the central question is: Which

combination of inclination of seat and backrest provide low shear forces? A biomechanical model was developed which predicts these combinations. The instrument described in chapter 2 was used for verification of the model on ten healthy subjects.
It was concluded that, when a small shear force is accepted, a fixed inner angle between seat and backrest must be chosen between 90° and 95°. For tiltable chairs this holds for backrest inclinations to 70°.
For the inclinations of seat and backrest in beds a parabolic relationship was found. The maximum seat inclination of 20° is found at a backrest inclination of 50°. When lying with bent knees and equal inclination of upper and lower leg, the model predicts a shear force on the buttocks that shoves the person into the bed. This holds for every combination of seat and backrest inclination.
Pressure is measured more easily than shear and is therefore the more documented of the two in literature. The preceding chapters have dealt with shear as a factor in the development of decubitus, the following relates to pressure.
A great variety of wheelchair cushions are produced with the aim of reducing pressure. In most studies, dealing with the evaluation of wheelchair cushions, the pressure is measured under the ischial tuberosities in a *static* situation and in only one posture. Wheelchair users, however, interact with their environment and during the day they drive on different ground surfaces in different postures which leads to continual pressure variations. In **chapter 6** the pressure under the buttocks of subjects with spinal cord injuries was measured continuously during wheelchair driving, while they were sitting on a foam cushion (thickness 5 cm) or a Jay Active™ cushion. The results of dynamic pressure measurements correlate with three classes of tissue response under the buttocks: severe, moderate and mild. A 5 cm thick foam cushion only produced severe responses, while a Jay Active™ cushion showed mild responses in most cases.
Finally dynamic aspects of sitting in an office chair are dealt with in this thesis. The purpose of a good chair is to prevent continuous overload of the lumbar spine, however, in most studies the lower lumbar curvature is measured in a static situation.
In **chapter 7** the lower lumbar curvature was measured continuously during a 55 minute reading task when seated on an office chair with the

seat tilted forward and backward and on a balans chair. On all these chairs the lumbar curvature becomes more kyphotic (flattened) compared to when the subjects is standing. The mean lumbar curvature during a 55 minute reading task is significantly more lordotic on the balans chair compared to an office chair with a forward tilted seat.
There is no significant difference in mean lumbar curvature during a 55 minute reading task when sitting on a balans chair or on an office chair with a backward tilted seat.

The studies mentioned above primarily deal initially with the *measurement* of physical quantities. Which values of the quantities measured are decisive for good comfort and prevention of health problems, is a question which must be elaborated upon in further studies.

Discussie en samenvatting

Discussie en samenvatting

In de moderne westerse samenleving neemt het aandeel van zitten op stoelen in werk en vrije tijd toe. Zorg daaromtrent betreft de enorme omvang van rugklachten; 80% van de volwassen Amerikanen heeft gedurende het leven tenminste eenmaal problemen met de lage rug. Er wordt aangenomen dat de vormgeving van meubilair een belangrijke invloed heeft op de etiologie. Alhoewel het onderwerp eenvoudig lijkt, blijkt het selecteren van geschikt meubilair, zowel voor thuis als voor professionele toepassing, nog steeds problemen op te leveren. Wij schrijven dit toe aan een gebrek aan bestaande kennis over ontwerpcriteria, en voegen daaraan toe dat veel van deze kennis bovendien nog onderwerp van discussie is en vaak berust op onvoldoende validering. Wij stellen ook vast dat nog vele vraagstellingen niet of onvoldoende zijn onderzocht, hetgeen te maken heeft met het ontbreken van apparatuur voor het in vivo meten van complexe grootheden.
Met dit voor ogen zijn in de vakgroep Biomedische Natuurkunde en Technologie aan de Erasmus Universiteit Rotterdam nieuwe laboratorium opstellingen en een ambulante houdings- en bewegings registratie set ontwikkeld. Deze computer geïntegreerde apparatuur maakt het mogelijk een groot aantal data over complexe fysische grootheden te verzamelen. Langs deze weg is in deze studie ingegaan op belastingverdeling, met name lokale druk en afschuifkracht, en de curvatuur van de lumbale wervelkolom in relatie tot individuele karakteristieken, vorm en afmetingen van meubilair, en materiaaleigenschappen van lichaamsondersteuningsvlakken.
Als zodanig betreft deze studie een aantal deelprojecten op het gebied van lichaamsondersteuning die elk nieuwe inzichten hebben opgeleverd dan wel veronderstellingen uit de literatuur hebben getoetst en gekwantificeerd. Deze resultaten worden in het kort genoemd in de navolgende samenvatting van de verschillende hoofdstukken.

Kennis van de belastingverdeling tussen de mens en stoelen of bedden geeft inzicht in de belasting van spieren, botten en zacht weefsel.

In **hoofdstuk 2** wordt een nieuw instrument, de meetstoel, geïntroduceerd waarmee de krachten op drie verschillende lichaamsondersteuningsvlakken (rugvlak, zitvlak en voetvlak), in componenten loodrecht op het vlak en in langsrichting van het vlak, kunnen worden gemeten. Krachten werden geregistreerd tijdens het zitten op een configuratie van vlakken waarmee een bed werd gesimuleerd met het rugvlak onder een hoek van 45° en de matras horizontaal. Metingen van ondersteuningskrachten aan gezonde proefpersonen (gemiddelde massa 77, SD= 11 kg) konden worden verricht met een nauwkeurigheid van ± 10 N. In bed bedroeg de afschuifkracht op het zitvlak gemiddeld 97 N.
Nadat de afschuifkracht op het zitvlak was gekwantificeerd, concentreerde de volgende deelstudie zich op de verdeling van de afschuifkracht over het zitvlak.
Afschuifkracht wordt wel gezien als een belangrijke factor in het ontstaan van decubitus. In **hoofdstuk 3** wordt een nieuwe sensor geïntroduceerd, waarmee de locale afschuifkracht op vervormbare oppervlakken gemeten kan worden zonder het fenomeen afschuiving te verstoren. De sensor werd gevalideerd in een meting met 10 gezonde jonge proefpersonen, door de som van de locaal met de sensor gemeten afschuifkrachten te vergelijken met de totale krachten gemeten op de meetstoel.
Tijdens zitten op een opvouwbare rolstoel werd een afschuifspanning van 4.8 kPa geregistreerd in de longitudinale richting en 8.5 kPa in transversale richting. Op de matras van een ziekenhuisbed werd een afschuifspanning van 5.6 kPa in longitudinale richting gemeten en 3.1 kPa in transversale richting.
Uit de literatuur was bekend dat, wanneer de afschuifspanning voldoende hoog is (9.9 kPa), de combinatie van druk en afschuiving bijzonder effectief is in het stremmen van de doorbloeding van de handpalm. Dit gegeven gaf aanleiding tot de vraag wat de invloed zou zijn op de doorbloeding van de laagste afschuifspanning die werd gemeten met de nieuwe sensor (3.1 kPa).
In **hoofdstuk 4** wordt transcutane zuurstofspanning genomen als een maat voor de doorbloeding. De zuurstofspanning werd gemeten op het sacrum van gezonde, jonge proefpersonen (gemiddelde leeftijd 25.5, SD= 3.4 jaar). De afsluitdruk werd gedefinieerd als die druk waarbij de zuurstofspanning daalt tot een niveau van 1.3 kPa; dit is het niveau

waarbij ischemie van de huid kan worden verwacht. Een gemiddelde afsluitdruk van 11.6 kPa werd gevonden zonder afschuifspanning, terwijl een significant lagere afsluitdruk van 8.7 kPa werd gevonden bij een afschuifspanning van 3.1 kPa. Er werd geen significante relatie gevonden tussen afsluitdruk enerzijds en systolische bloeddruk, diastolische bloeddruk, huiddikte ter hoogte van het sacrum, vetpercentage en zuurstofspanning in de onbelaste toestand anderzijds.
Uit het bovenstaande werd geconcludeerd dat de standen van rugvlak en zitvlak belangrijke ontwerpcriteria zijn. In **hoofdstuk 5** staat de vraag centraal welke hoekcombinaties van rug- en zitvlak een lage afschuifkracht geven. Een biomechanisch model werd ontwikkeld om deze combinaties te voorspellen. Het meetinstrument, dat is beschreven in hoofdstuk 2, werd gebruikt om de voorspellingen van het model te verifiëren bij 10 gezonde proefpersonen.
Geconcludeerd werd dat, wanneer een kleine afschuifkracht kon worden toegelaten, een vaste hoek tussen rug- en zitvlak tussen 90° en 95° kan worden gekozen. Dit geldt voor kantelbare stoelen wanneer de rughoek niet verder kan kantelen dan 70°.
Voor de hoeken tussen zit- en rugvlak van bedden werd een parabolische relatie gevonden. De grootste zithoek van 20° wordt gevonden bij een rughoek van 50°. Bij een lighouding met gebogen knieën en gelijke helling van boven- en onderbeen voorspelt het model een afschuifkracht op het zitvlak die de persoon het bed in schuift. Dit geldt voor elke combinatie van zit- en rughoek.
In de voorgaande hoofdstukken lag de nadruk op afschuifkracht als een factor in het ontstaan van decubitus. Omdat het veel eenvoudiger is druk te meten dan afschuiving, is druk beter gedocumenteerd in de literatuur. Om de druk te reduceren wordt een groot aantal verschillende rolstoelkussens gefabriceerd. In de meeste studies waarin rolstoelkussens worden geëvalueerd wordt de druk onder de zitbeenknobbels gemeten in een *statische* situatie in één zithouding. Maar rolstoelgebruikers hebben interactie met hun omgeving en rijden over verschillende oppervlakken in verschillende houdingen gedurende de dag, hetgeen leidt tot continue drukvariatie. In **hoofdstuk 6** wordt de druk onder de zitbeenknobbels van 9 dwarslaesie patiënten continue gemeten tijdens het rijden in een rolstoel, terwijl zij zitten op een standaard schuimrubber kussen (5 cm dik) en op een Jay Active™ kussen. De resultaten van dynamische

drukmetingen kunnen worden gecorreleerd met drie klassen weefselrespons: ernstig, matig en mild. Het schuimrubber kussen leverde in alle gevallen ernstige weefselrespons, terwijl het Jay Active™ kussen in de meeste gevallen een milde weefselrespons opleverde.
Tenslotte worden in dit proefschrift de dynamische aspecten van zitten op een kantoorstoel belicht. Een goede kantoorstoel moet continue overbelasting van de lumbale rug voorkomen. In de meeste studies wordt de lumbale curvatuur echter in een statische situatie gemeten.
In **hoofdstuk 7** werd tijdens het uitvoeren van een leestaak gedurende de 55 minuten de lumbale curvatuur continu gemeten, terwijl de proefpersonen afwisselend zaten op een conventionele kantoorstoel met een voorover- en achterover gekanteld zitvlak en op een balans stoel. Op elk van deze stoelen werd de lumbale wervelkolom rechter in vergelijking met de vorm bij staan. De gemiddelde lumbale kromming bleek significant meer lordotisch te zijn tijdens zitten op een balans stoel in vergelijking met zitten op een conventionele kantoorstoel met vooroverhellend zitvlak. Er was geen significant verschil in lumbale curvatuur bij vergelijking van een balans stoel met een conventionele kantoorstoel met achteroverhellend zitvlak.

De bovenstaande deelstudies betreffen primair het *meten* van fysische grootheden. De vraag welke waarden bepalend zijn voor comfort en voor preventie van gezondheidsklachten, moet nader worden onderzocht.

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Tot slot

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Curriculum Vitae

Richard Goossens was born on November 4th, 1965, in Rotterdam, The Netherlands. In 1984 he finished gymnasium-ß. In 1984 he was matriculated into Delft University of Technology. After four and a half year he graduated in mechanical engineering with specialization robotics. From June 1989 until December 1993, he was employed by the Department of Biomedical Physics and Technology at the Erasmus University of Rotterdam, where he conducted his study on the biomechanics of body support.