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SHEAR STRESS MEASURED ON BEDS AND WHEELCHAIRS

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ABSTRACT. Local shear is understood to be one of the principal risk factors for the development of pressure sores. There is a need for a small deformable sensor that can measure the shear force between skin and deformable materials without disturbing the shear phenomenon. In the present study a new shear sensor is introduced with a contact area of 4.05 cm^2. A series of validation experiments was performed with ten healthy young subjects. 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 while sitting in 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, while in sitting position in bed.

Key words: bed, decubitus, shear, wheelchairs.

INTRODUCTION

When forced to lie down or sit for long periods some people run the risk of developing 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 (6, 12, 16, 17, 20, 21). Some of these authors believe that paraplegia or denervation is an extra factor in the occurrence of pressure sores. Daniel et al. (5) suggest that because of atrophy of tissue the padding around the bony prominences is reduced, resulting in higher interface pressure. Research on animals revealed an inverse relationship between intensity and duration of applied pressure. To explain this inverse relationship Reddy et al. (23) 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 ischaemic 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 up in bed, the shear force acting on the buttocks can be influenced by changing the inclination angle of the upper legs (26), or, secondly, by changing the material and profile of the seat or the mattress. For example, sitting on a wooden surface produces higher pressure than sitting on a soft foam cushion. In the most research the influence of the material on pressure is evaluated (7, 14, 15, 18, 19, 22, 27).

In 1958 it was Reichel (24) who started to focus the attention on shear force as an important component of mechanical load. Shear force is defined as a force that acts *parallel* to a surface (whereas pressure acts *perpendicular* to a surface). Until now few articles have been published on this subject, because shear is extremely difficult to measure. Dinsdale (6) 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. (2–4) 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. Goossens et al. (9) also studied the influence of shear on blood flow. A shear stress of 3.1 kPa and a significant influence on the reduction of blood flow on the sacrum of healthy subjects.

The shear sensors used in previous studies, however, could only be used on hard surfaces, thus confirming the need 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 a sitting and lying position. The shear stress was measured in a static situation. 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. This maximum shear force just before sliding occurs is defined by:

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

where f is the 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 the literature. The measurements of Garber et al. (7), Krouskop et al. (19), and Seymour & Lacefield (25) 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 (1). 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 presumed 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 (26) showed that the angle of the backrest and the seat must be coupled by means of a biomechanical model. So it can be presumed that the local shear stress can be affected by the choice of material and by tilting the seat and the angle of the backrest.

METHODS

When developing a shear sensor, two aspects are of major importance. Firstly, there is the surface of the sensor. When this is too smooth, the correct magnitude of the shear force might not be detected, because the sensor slides over the surface. Secondly, the dimensions of the shear sensor have to be considered. 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 the study by Bennett et al. (2). 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 (Fig. 1).

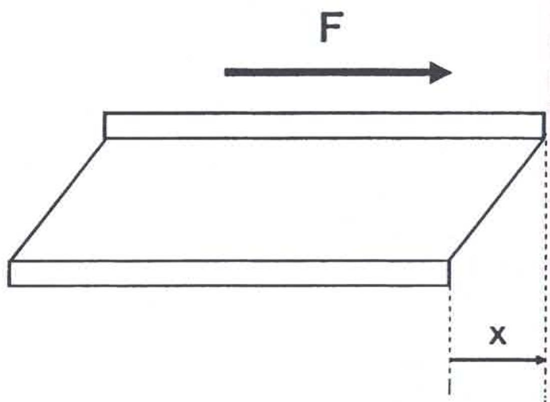

Fig. 1. The principle behind the new sensor. A force implies a certain displacement (x) causing a change in capacitance.

The prototype sensor was developed in close cooperation with Delft University of Technology and Erasmus University Rotterdam. The sensor is based on a capacitive principle by Heerens (13). He used a principle that was developed more than a century ago by Kelvin. 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 environment. Kelvin found a 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, thus allowing very small changes in capacitance to be measured very accurately. In order to measure the displacement in two directions, four capacitors are needed.

The sensors are $27 \times 15 \times 3.5$ mm, in size, and thus the contact area is $4.05\ cm^2$. Six of these sensors were glued onto a thin slice of rubber (thickness 1 mm) to keep their relative position the same.

I. Measurements to validate the sensor

To study the influence of the shear sensor on the force measurements, measurements were conducted with and without shear sensor underneath the buttocks of the subject. Therefore, in this test, shear on the seat was measured in two ways: resultant shear force with the aid of a force plate (in this article called "global shear force") and local shear force acting on the surface of the new shear sensor (in this article called "local shear force"). The global shear force was measured with a force plate that is part of a measuring chair (11). 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 on the whole surface, the seat was divided into 4 rows and 8 columns. The sensor mat was placed on all 32 positions. The shear stress measured by the sensor was converted into shear force by multiplying by the area of the sensor ($4.05\,cm^2$). The shear force was then resolved in the longitudinal and transversal directions 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 four 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 (11) 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, and when the seat is angled at 8° the shear force becomes smaller than 5 N (in healthy subjects).

In the present wheelchair test, local shear stress under the left buttock was studied in four 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 the 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 in sitting position in bed with a trunk angle of 45° and legs horizontal, a shear force acts on the buttocks (10). 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 seat varied from 0° to 20° in steps of 5°. In each subject five measurements were taken. The maximum value of each measurement was used for statistics.

For subsequent tests the sequence of the situations was randomized. Ten healthy subjects were measured (mass 68 (s.d. 14) kg, length 177 (s.d. 10) cm, age 24 (s.d. 3) years). Between every situation the subjects stood up in order to allow adjustments to the measuring chair to be made. To ascertain equal coefficients all subjects wore a pair of cotton trousers, as used in surgery. The measurements in test I were performed in the longitudinal direction only and in tests II and III in both the transversal and longitudinal directions.

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:

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

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

In tests 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

A typical distribution of the shear stress in the four situations can be seen in Fig. 2.

Table I shows the results of the sum of the local shear forces and the resultant shear force.

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 presented in Table II.

An 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 in shear stress between wood and gel with a seat angle of 0° ($p = 0.01$) as well as with 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 III).

Analyses of variances showed no significant

Table I. *Mean and s.d. (in parentheses) of the local and resultant forces as measured on the seat from ten subjects*

	Wood Forward (10°) [N]	Backward (−10°) [N]	Foam 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)

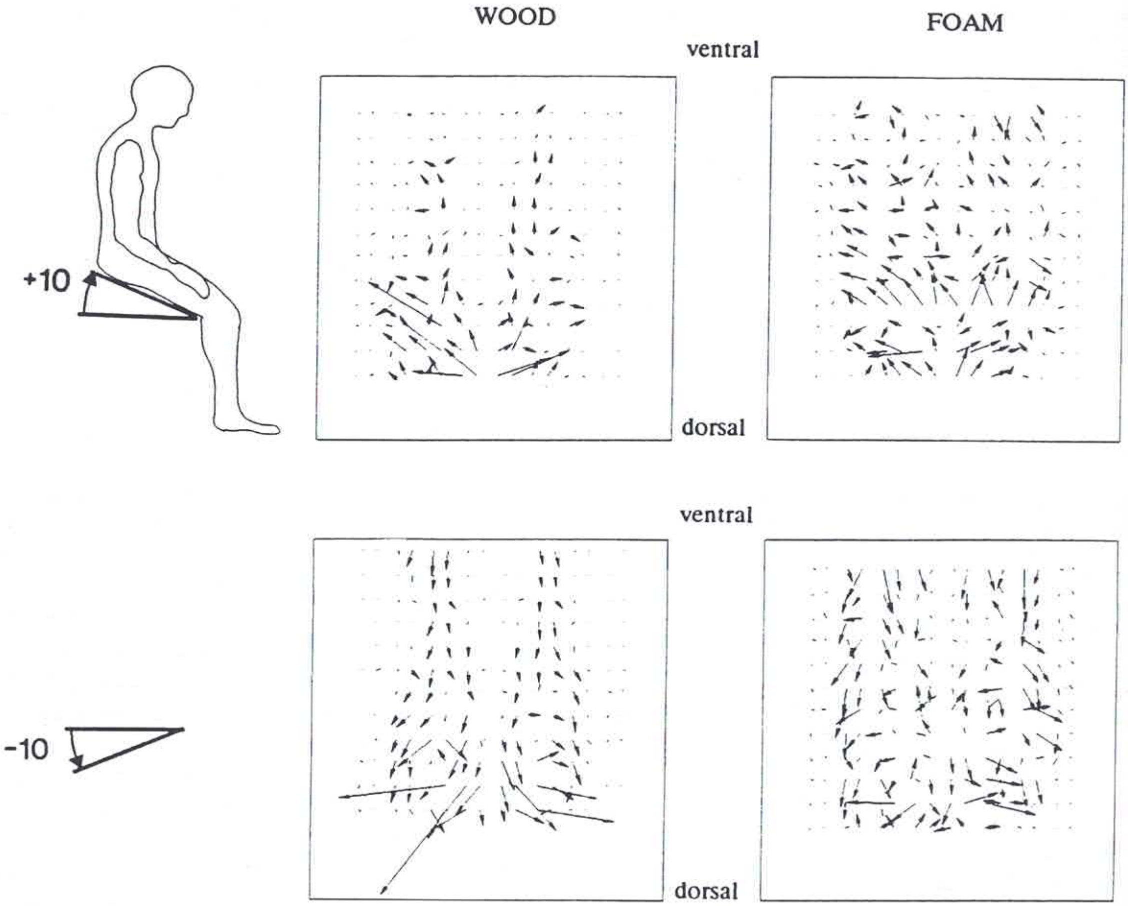

Fig. 2. Distribution of the local shear on a forward- and backward-tilted seat of wood and gel recorded from one subject. As can be seen, the regions of highest shear stress are at the ischial tuberosities (the regions of highest pressure).

difference for the different seat angles, either in the longitudinal or in the transversal direction.

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 the literature is that local shear plays a role when it is above 3.1 kPa (9). Compared to a healthy population, Bennett et al. (3) found smaller blood volume rates and higher shear forces in paraplegics and hospitalized geriatric patients.

It is assumed that when an unfavourable effect of shear stress can be measured in healthy young subjects, the effect for the hospitalized geriatric and paraplegic population will be even worse.

Table II. *The mean of the maximum local shear force under the left tuberosity of ten subjects in kPa (s.d. in parentheses)*

	No cushion Longitudinal [kPa]	Transversal [kPa]	Gel cushion 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)

Table III. *The mean and (s.d. in parentheses) in kPa of the maximum local shear force on the sacrum of ten healthy subjects lying in a hospital bed with different seat angles*

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

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 2 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 presumed 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 with a soft covering. However, when the seat is tilted forward, the sum of the local shear forces is significantly 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 with those found in the literature. Gilsdorf et al. (8) found a resultant shear force on the seat of 27 N on a hard surface and 50 N on a ROHO cushion. Stumbaum (28) 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 position 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 (Table I). 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 position, because of leg support that prevents the subject from sliding. This implies 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, s.d. 11 N).

In the second test, shear stresses on soft and hard material were compared. The significantly larger shear stress in the transversal direction on soft material can be explained by larger material deformation.

In the third test the relationship between local shear stress and seat angle was studied. Previous measurements showed that the resultant shear force on the seat tends to decrease with increasing seat angle. 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 II) or a bed (Table III). A further study on one subject showed that when the subject remained seated 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 to 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 disappeared. Because in this study the subject stood up between every measurement, the small change in shear force resulting from the change in angle was not measurable. Still, the local shear forces presented in this study are likely to have a practical value as an average for different situations; which can be compared with the permissible limit.

Local shear on a hard seat was measured by Bennett et al. (3). He, too, found that the local shear stress is independent of the seat angle. The device used by Bennett et al. was based on strain gauges and the measurements were taken on a geriatric and a healthy population. Shear was measured with a sensor with an area of 0.20 cm^2 (20 times smaller than the sensor used in this study), the position being 9 cm lateral of the seat centre 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 in the geriatric population was almost 3 times as high as that found in the healthy population (average 2.6 kPa vs 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 cannot be determined, possibly because of the influence of body posture.
- In a standard foldable wheelchair the average local shear stress at the tuberosities equals 4.8 kPa in the longitudinal direction and 8.5 kPa in the transversal direction.
- When in sitting position in a hospital bed the average 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, for example head and arm movements.

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