

Comparative effects of posture on pressure and shear at the body-seat interface

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Abstract—This study considers the effects of seated posture and body orientation on the pressure-distribution and surface shear (tangential) forces acting at the body-seat interface. Nine postures typically assumed by wheelchair users were studied. Comparisons were made within and between two study groups, made up of 12 subjects with spinal cord injuries (SCI) and 10 nondisabled subjects. Both interface pressure and the surface shear were measured simultaneously in each of nine reproducible, seated postures. The same seat cushion was used for all trials. The Oxford Pressure Monitor, a pneumatic cell device, was used to measure and record the interface pressures. Instrumentation for measuring and recording the surface shear force was constructed specifically for the study. Analysis consisted of statistically comparing changes in pressure values and shear forces derived from eight sitting postures with reference to values recorded in a defined neutral sitting posture. The pressure-distribution findings suggest that in the postures studied SCI subjects have maximum pressures that are higher than nondisabled subjects in all postures, ranging from 6% to 46% depending on the posture. Maximum pressures can be reduced by postural changes: forward flexion to 50°, -9%; backrest recline to 120°, -12%; and, full body tilt, -11%. On average, the SCI group members have peak pressure gradients (PPG) that are 1.5 to 2.5 greater than the nondisabled group. The maximum reduction in PPG occurred at backrest recline of 120°, -18%. Tangential shear force acts at the body-seat interface in all nine postures studied. Extrapolation of results suggests that full-body tilt to approximately 25° reduces the surface shear force to near zero. In contrast, a backrest-

only recline of 20° causes a 25% increase in the surface shear force. These results suggest that caution must be taken when using nondisabled subjects as surrogates for people with SCI because of the inherent differences between the groups. Also, researchers and clinicians should recognize that posture and body orientation in space are additional variables that can have a profound effect on the interaction between a seated person and his or her supporting surface.

Key words: *body-seat interface, pressure distribution, spinal cord injury, surface shear force.*

INTRODUCTION

Studies conducted over the past several decades have confirmed that the incidence and related costs of pressure-sore treatment remain a major health problem (1,2,3). Support of body weight while lying, sitting, or standing necessitates the transmission of internal stabilizing forces via the supporting tissues to external support surfaces. This results in tissue-distorting forces being transmitted to underlying soft tissues. Excessive or prolonged application of distorting mechanical forces can result in vascular occlusion, ischemia with eventual necrosis and the onset of a "pressure sore" (4,5). The onset usually occurs in the deeper tissues and then migrates toward the surface (6,7).

Both clinical and research evidence, although conflicting and inconclusive in several areas, are strongly biased toward excessive and/or prolonged application of surface pressure as being the predominant causative factors in pressure-sore formation.

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Although at least one investigator has questioned the primacy of pressure as a causal factor in pressure-sore formation (8), the fact remains that there is a dearth of research evidence suggesting any more dominant factor than interface pressure.

There is also compelling evidence that factors in addition to pressure are contributors and must also be considered when attempting to fully understand the pressure-sore phenomenon. Studies have implicated factors such as shear stress (9), impact loading of tissue (5), elevated temperature and humidity (10,11), age, nutritional status, general health, activity level (12), deformity, posture and postural change (13,14), body stature (15), and psychological deficits. Crenshaw and Vistnes (16), in their review of pressure sore research, effectively highlighted the studies and current thinking on the etiology and related factors that can predispose a person to pressure-sore formation.

Numerous research techniques and devices have been developed in an attempt to quantify the interface pressure. Mooney (17) described the design of a flexible polyvinylchloride (PVC) airtight pneumatic cell with follow-up clinical trials. Fernie (18) did an exhaustive analysis of the interface technology used before 1973. Ferguson-Pell (19) outlined the design criteria for an ideal interface-pressure-measurement transducer for measuring pressure differences across nonuniform surfaces. Garber and Krouskop (15) described the design and clinical use of a 12 × 12-matrix transducer device, commercially termed the Texas Interface Pressure Evaluator (TIPE). Bader (20) and Bader and Hawkins (21) reported on the development of the Oxford Pressure Monitoring (OPM) device that built on the previous work of investigators such as O'Leary and Lyddy (22), who first described the principle upon which the OPM is based.

The focus of research and clinical efforts with the cited measurement tools has been to quantify the interface pressure between a seated person and the supporting substructure, usually a seat cushion in a wheelchair. Efforts have been made to establish absolute threshold values above which seated persons are likely to be at risk of pressure-sore formation. Other applications have attempted to make comparisons between various supporting materials and related commercial products to guide improved product selection. Studies by Reddy *et al.* (23) and Guthrie (24) have emphasized the difficulty

in obtaining and accurately interpreting absolute pressure values, especially when used in nonplanar environments consisting of materials with compliance properties significantly different from human soft tissue. Reddy *et al.* (23) came to the conclusion that comparative values rather than absolute values are more likely to be meaningful, especially when the interface materials can be kept constant and the materials have mechanical properties approaching those of human buttock tissue.

Review of the research literature indicates that little attention has been given to the effects of posture and deformity on pressure distribution at the seat interface or to how those effects might vary between nondisabled and people with a spinal cord lesion. Zackarkow (14) strongly emphasized the importance of these factors and the need for research verification of clinical observations.

Over the years, both researchers and clinicians have indicated a concern about the effects of shear stress acting at the body-support interface. Guttmann (25) emphasized the importance of distinguishing between purely vertical pressure and shear stress: "Shear stress is much more disastrous for it cuts off larger areas from their vascular supply." Reichel (26) expressed an opinion that raising the head of a hospital bed by even a few inches was capable of producing sufficient shear force over the sacral area to deprive large tissue areas of blood supply. Bennett *et al.* (9) came to the conclusion that combined pressure plus shear particularly promotes occlusion. The value of pressure necessary to produce occlusion can be nearly halved when accompanied by sufficient shear. Other investigators have also stressed that tissue deformation is a major etiological factor in ulcer formation (8,27,28). Chow and Odell (29) made the point that shear stress is involved in uniaxial pressure, localized pressure, and any nonuniform pressure distribution or pressure that causes tissue distortion.

In addition to shear stresses induced by the normal pressure, there are also shear stresses caused by forces that are acting tangential to the support surface, that is, friction forces. Very little progress has been made toward the measurement and use of frictional shear force information in clinical decision making (16). This is due, in part, to the complexity of the interface shear-stress phenomenon when viewed in its entirety. However, if one differentiates between the types of shear stresses, normally in-

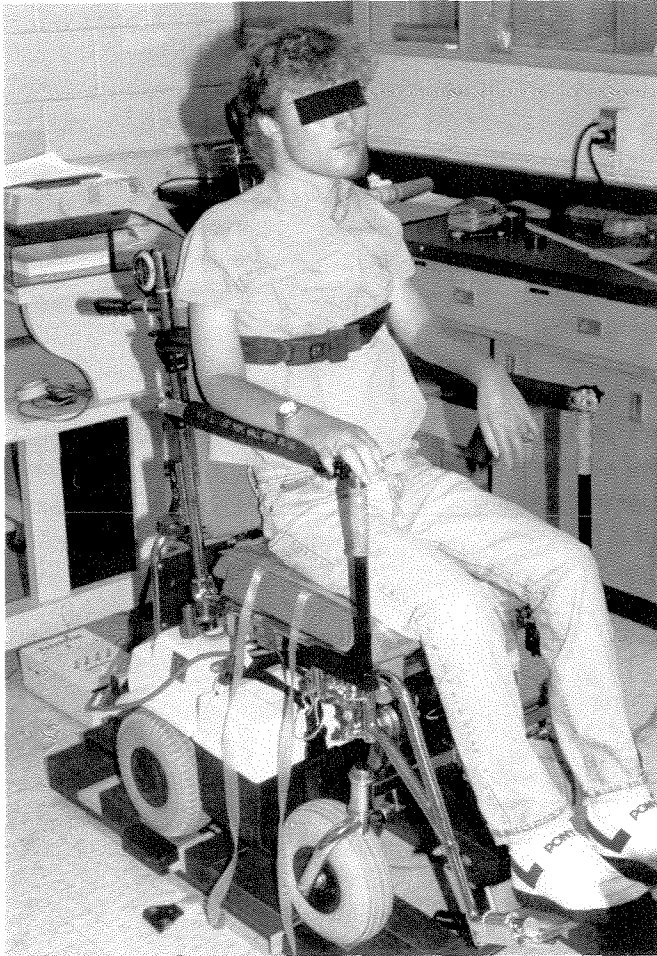


Figure 1. Disabled subject positioned on the BPC in neutral position (arms are placed on lap during pressure- and shear-recording sessions).

duced and tangentially induced, one can conceive of practical methods to measure the simpler tangentially induced shear (TIS) forces. Adoption of TIS measurements as a practical tool could provide clinicians with information about a variable, in addition to the pressure variable, that should be factored into clinical decision-making processes. However, it should be stated that the manner in which the shear- and normal-force components interact within the tissues will not be illuminated by this rather direct and simplistic approach.

In summary, this study has investigated the effects of changes in sitting postures and body orientation in space on interface pressure distribution and on TIS that acts at the body-seat interface.

OBJECTIVES

The objectives of this study were: 1) to identify pressure-distribution differences that may exist within and between the nondisabled and spinal cord injured (SCI) populations and to determine how the distributions may be affected as a result of deformity and/or alterations in sitting postures; and, 2) to investigate the manner in which TIS forces change with body posture, both within and between an SCI group and a nondisabled control group.

METHOD AND MATERIALS

Ten nondisabled and 12 SCI subjects participated in the study. The nondisabled group consisted of six males and four females with a mean age of 39.3 years and a range from 28 to 57 years. Mean body weight was 68.6 kg with a range of from 50 to 95.5 kg. All subjects in this group reported no previous impairments of spinal or pelvic movement.

All 12 SCI subjects were diagnosed as having complete lesions of their spinal cords for at least 5 years. The sample contained 10 males and 2 females: 7 paraplegic and 5 quadriplegic. Mean age was 40.9 years with a range of 25–66 years. Mean body weight was 59.8 kg with a range of 39–74.2 kg. The mean number of years since injury was 19.5 with a range of 6–54 years. In general, the subjects in the SCI group were all active individuals living in the community who regularly use a wheelchair.

Nine typical wheelchair sitting postures, as defined below, were studied. A specially designed body-positioning chair (BPC), using a Fortress model FS655 (Fortress, Inc.) with a powered recliner as the base, was used to obtain consistency of seated postures between subjects and between trials (**Figure 1**). The seat cushion used was constructed of three layers of 25-mm (1 in) Sunmate(b) foam (type, soft blue, 5 lb/ft³).

Two OPMs were used to measure the pressure values under the buttock area. Four, 3×4 cell-matrix transducers (48 cells) were fixed to the seat surface so that their positions relative to a reference axis would remain constant. The transducer array was placed so that the ischial tuberosities of all subjects would fall on the back two arrays (24 cells). The front two transducer arrays were used only to ensure that the maximum

pressure values actually occurred within the back two arrays.

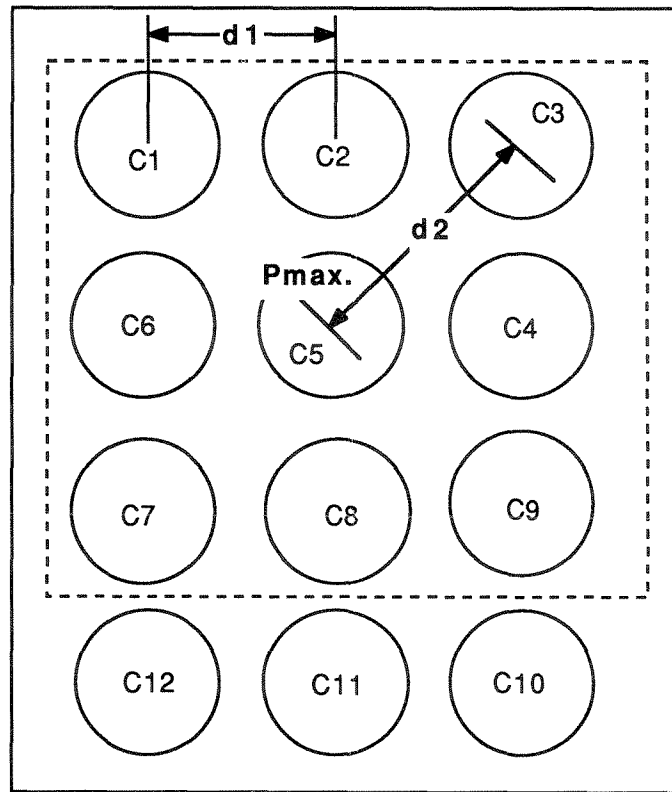
The subjects were first positioned in the P1M (neutral) position on the BPC, and pressure recordings were taken in the P1M and all remaining eight standard positions (P1L to P7). Subjects were given a 15-minute free-movement period before each recording series to allow the seat foam and measurement equipment to reach steady state. Maximum pressure values, location of maximum values, mean average pressure values, and mean peak pressure gradients were all computed from the recorded data. **Figure 2** defines the terms maximum, average, and peak pressure gradients. The subjects were not removed from the BPC between changes in body postures, thereby minimizing errors that could result from inaccuracies associated with reseating subjects (20). Successive pressure recordings were taken in one posture in order to conduct repeatability and variability checks on the pressure measurements. Calibration of the OPMs was done at the beginning of each trial and again at the end of those trials in which any comparatively high or low values were observed.

The forces acting tangentially to the seat surface (TIS) were measured in both the fore and aft directions. The seat surface (foam plus plywood substrate) was mounted on an aluminum substructure within the BPC. The seat was constrained by the load cells mounted at each end of the seat substructure so the seat would move not freely in the fore and aft directions. As a result, the load cells measured the horizontal reaction forces between the seat structure and the wheelchair frame, including the backrest, that tended to move the seat in the fore and aft directions. These forces were defined as the TIS or friction forces acting at the seat-body interface.

The TIS was measured simultaneously with the interface pressure in each of the nine standard sitting postures defined below. Mean values and standard deviations for both groups were computed and statistically analyzed. Significance was declared at the $p = 0.05$ level. The Student's t -test was used to test the paired and unpaired samples because of its increased sensitivity to errors associated with small samples (30).

Instrumentation and calibration

Pressure Measurement. The OPM was devel-



$$\text{Average Pressure} = \frac{\text{Sum}(C1...C12)}{12}$$

$$\text{Maximum Pressure} = P_{\text{max}}(C1...C12)$$

$$\text{Peak Gradient} = \frac{P_{\text{max}} - P_{\text{min}}(C1...C9)}{d1 \text{ or } d2}$$

Figure 2. Definitions of average, maximum, and peak pressure gradients.

oped at the Oxford Orthopedic Engineering Centre, Nuffield Orthopedic Centre, Oxford, England (21). The OPM, which works on the principle first described by O'Leary and Lyddy (22), consists of three basic components; the pneumatic transducer array, the monitor, and the calibration jig.

The transducers are supplied in either a 3×4 matrix configuration, a 1×6 matrix, or as single cells. An important feature of OPM development is that there are no electrical contacts necessary within the pneumatic transducer itself. Also, a transducer cell is only inflated when a pressure reading is taking place, thereby minimizing the perturbation effect on surrounding body-contact areas. The 3×4 matrix

configuration used in this study is comprised of 12 20-mm-diameter cells mounted on 28-mm centers. This matrix provides an effective measurement area of approximately 8,000 mm². Each cell in the matrix is fed by a small-diameter flexible tube that plugs into the monitor manifold switch. The thickness of the deflated matrix is approximately 2-3 mm. The OPM is powered by a 12 V (6-Ah) rechargeable Ni-Cd battery. The stated pressure range is from 0 to 300 mmHg, with a resolution of 1 mmHg.

The calibration jig is composed of three components: a pressure gauge, a double-sided vinyl bladder, and a slotted compression box. The procedure involves placing the cell transducer matrix between the two halves of the inflatable bladder. The bladder/transducer assembly is then placed in the compression box, and the bladder is inflated to 100 mmHg as indicated by an interconnected external pressure gauge. The manifold of the OPM is then rotated through the various cell settings, and any variation from the 100-mmHg calibration value is noted. A calibration setting on the side of the OPM allows adjustment of the recorded values to the 100-mmHg calibration value. This procedure permitted calibration of the two OPM units used to within ± 3 percent of the 100-mmHg calibration level across the cell matrix. This level of accuracy was achieved consistently throughout the recording session. Calibration data published by the developers (21) show a maximum deviation from linearity of 3 percent over the optimum range of 0-250 mmHg.

Rationale for the Selection of the Oxford Pressure Monitor. A review of previous work on qualifying pressure distribution revealed a wide selection of possible approaches. Each approach is not without its limitations. In general, miniature electronic transducers proved difficult to position accurately, are expensive, and are not readily available in matrix configurations that could be applied to the needs of this study. Also, reports of hysteresis problems, contamination by tangential forces, and electrical interference discouraged use of miniature electronic devices.

Preliminary trials were conducted with the TIPE device first developed at the Texas Institute for Rehabilitation Research in Houston, Texas (15). The device uses a 12 \times 12 pneumatic transducer matrix, which is attractive because of the large surface area it covers. However, the maximum pressure of 150 mmHg was found inadequate. Also,

consistent problems were encountered with the electrical contacts within the transducers. And finally, no calibration method is provided with the TIPE system, making it difficult to conduct routine calibrations.

Suggestions have been made that accuracy is improved when a single transducer cell is attached directly to the skin over an ischial tuberosity, rather than placing an array on the test surface. The radiographic results show that the ischial tuberosities shift approximately 4 cm during the complete range of movements studied. Therefore, the single-cell skin method is not appropriate for experiments in which large changes in trunk positions are to be studied.

Finally, 3 years of previous experience had shown the OPM to be a reliable and durable device that provides reasonably accurate results within known limitations. Linearity and variability tests have been done and published by the developers (20,21). The OPM is commercially available, and for this and the above reasons it was chosen as the most appropriate pressure-measurement system to use in the study. Furthermore, the fundamental approach of using relative rather than absolute values with the same seat cushion means that the errors associated with the OPM at least will be consistent and therefore largely self-canceling (23).

Shear Measurement. The load cells used were Sensortronics, model 60040, which is a low-profile bidirectional version that is produced in a variety of load capacities. The data-acquisition system used was a Metrabyte DAS-16 multifunctional high-speed analog-digital (A/D) I/O expansion board for the IBM PC computer. The front-end signal amplification and conditioning are done using a commercial unit from Action Instruments, model AP4051. The BASIC software program combined with the Metrabyte A/D board permitted calibration and adjustment of the baseline settings for the load cells before each run. Calibration of the load cells was done using a spring scale. A plot of the force-versus-voltage output indicated a linearity of approximately 2 percent within the operating range of the scale (0-25 lb, 0-11.3 kg). Tests were done to determine the friction force in the linear bearings, which in turn was used to adjust the recorded values of the load cells. Regression computations were necessary because the friction values varied slightly with body weight.

Definitions of standardized postures

The following definitions of the standard postures are illustrated in Figures 3, 4, and 5.

P1M-Position: Defined as the posture in the BPC with the pelvis placed as far posteriorly on the seat as possible, trunk and pelvis on the midline, and head in contact with the headrest. The seat surface is horizontal with the backrest reclined 10° to the vertical (100°). Arms are comfortably placed on the thighs, and footrests are adjusted to take approximately 10 percent of the body weight.

P1L-Trunk Bending-Left: Defined as the posture in

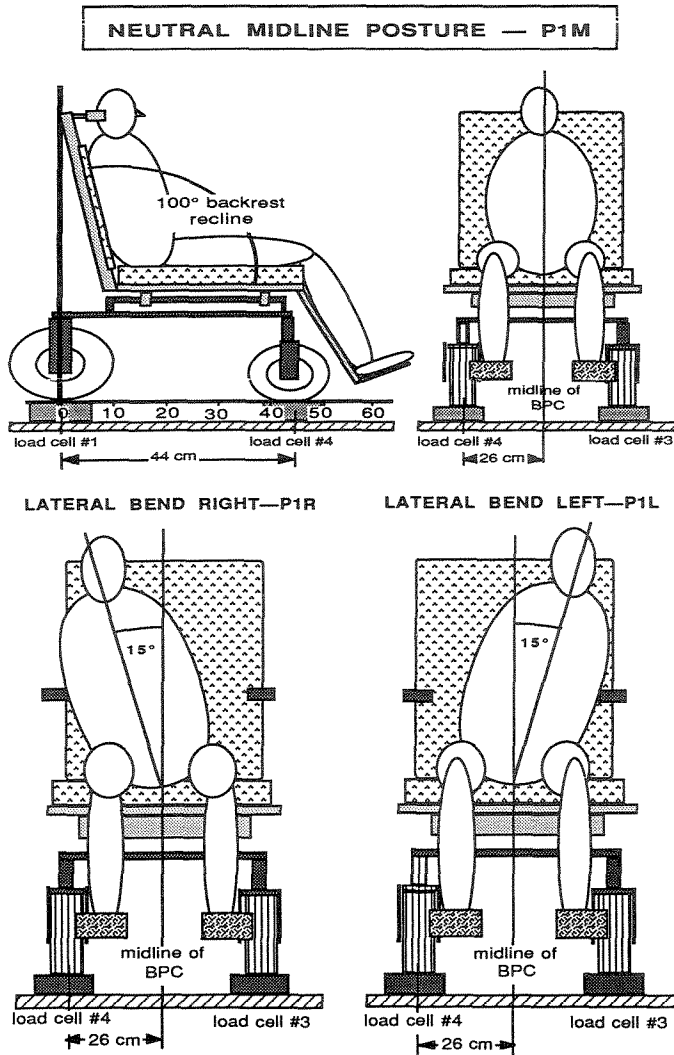


Figure 3.
Definitions of postures P1M, P1L, and P1R.

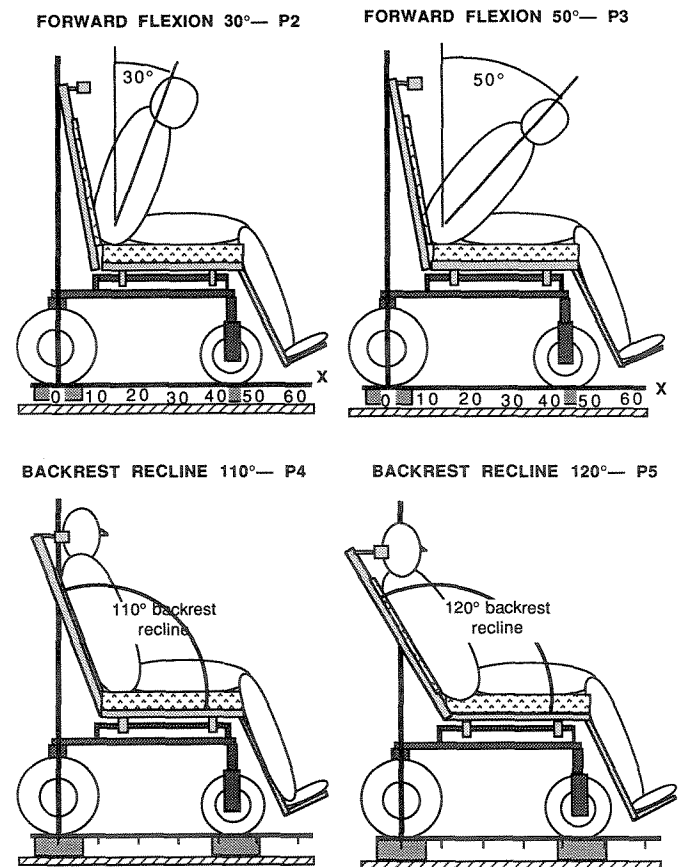


Figure 4.
Definitions of postures P2, P3, P4, and P5.

the BPC the same as P1M above, except that the trunk is flexed to the left until the left side contacts the elevated armrests (approximately 15° of lateral trunk flexion).

P1R-Trunk Bending-Right: Defined as the posture in the BPC the same as P1M, except the trunk is flexed to the right side.

P2-Forward Trunk Flexion 30° : Defined as the posture in the BPC the same as P1M, except the head and trunk are flexed forward 30° along the A/P midline.

P3-Forward Trunk Flexion 50° : Defined as the posture in the BPC the same as P1M, except the head and trunk are flexed forward 50° along the A/P midline.

P4-Back Recline 110° : Defined as the posture in the BPC the same as P1M, except the backrest is

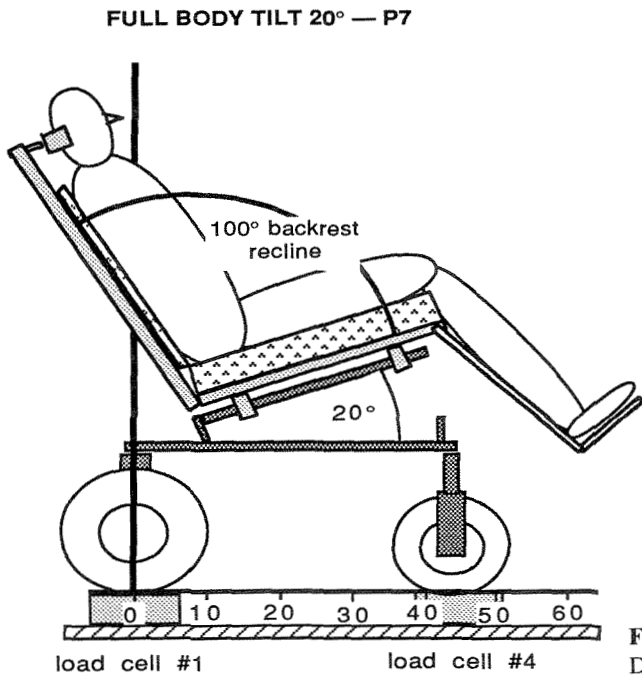
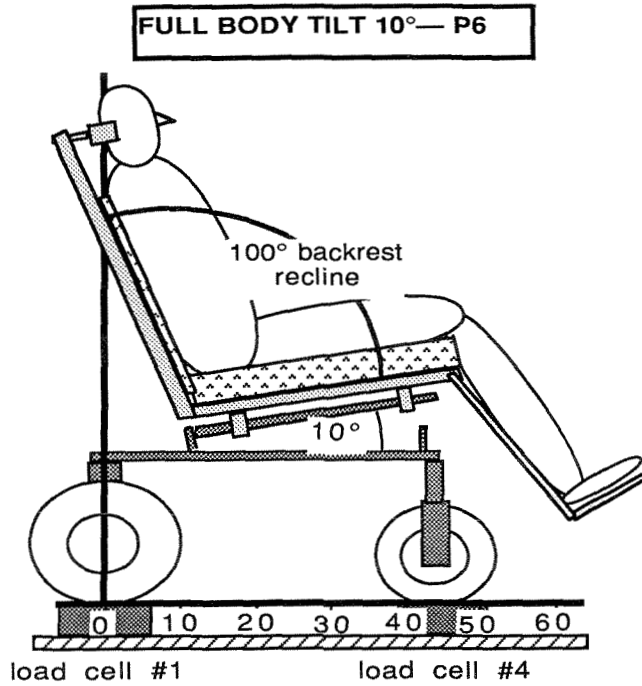


Figure 5. Definitions of postures P6 and P7.

reclined to the 110° position.

P5-Back Recline 120°: Defined as the posture in the BPC the same as P1M, except the backrest is reclined to the 120° position.

P6-Body Recline 10°: Defined as the posture in the

BPC the same as P1M, except the whole seat assembly (seat and back) is tilted 10° in space.

P7-Body Recline 20°: Defined as the posture in the BPC the same as P1M, except the whole seat assembly (seat and back) is tilted 20° in space.

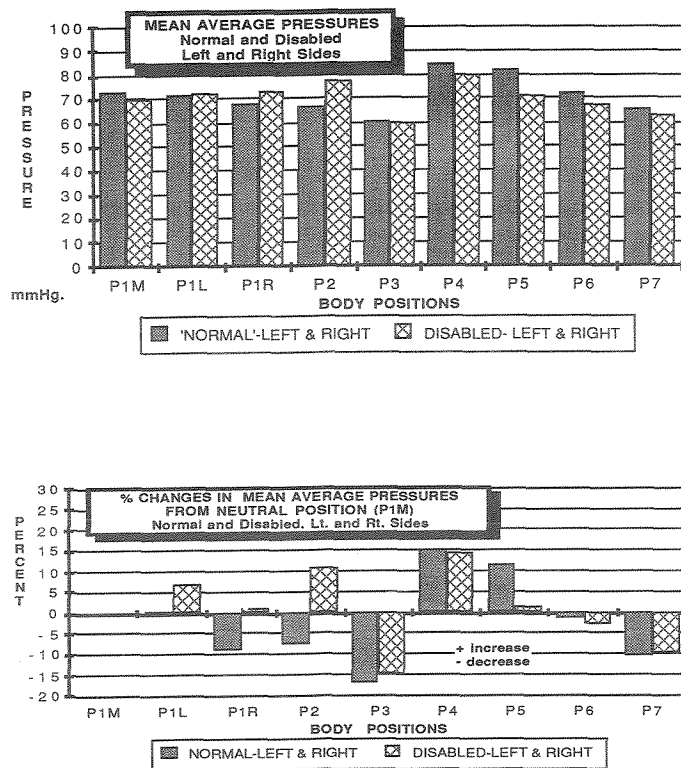


Figure 6. Results of pressure measurements averaged over 24 transducers. Upper plot shows the average values for both groups. Lower plot shows the percentage of change in the values from those in the neutral posture (P1M) as a result of changes in posture.

RESULTS

Pressure study

Figures 6, 7, and 8 graphically portray the results of the average, maximum, and peak pressure gradients for each study group. These results may be summarized as follows:

1. Average pressure distribution between populations is affected insignificantly by alterations of sitting posture. Forward flexion of the trunk causes the largest decrease within both groups from the average pressures in the neutral position. On average, a reduction of approximately 15 percent from the values in the neutral position occurs at 50° of trunk flexion (P3). However, the SCI group experiences an increase of approximately 10 percent at 30° of forward flexion (P2) before a reduction begins to take place.

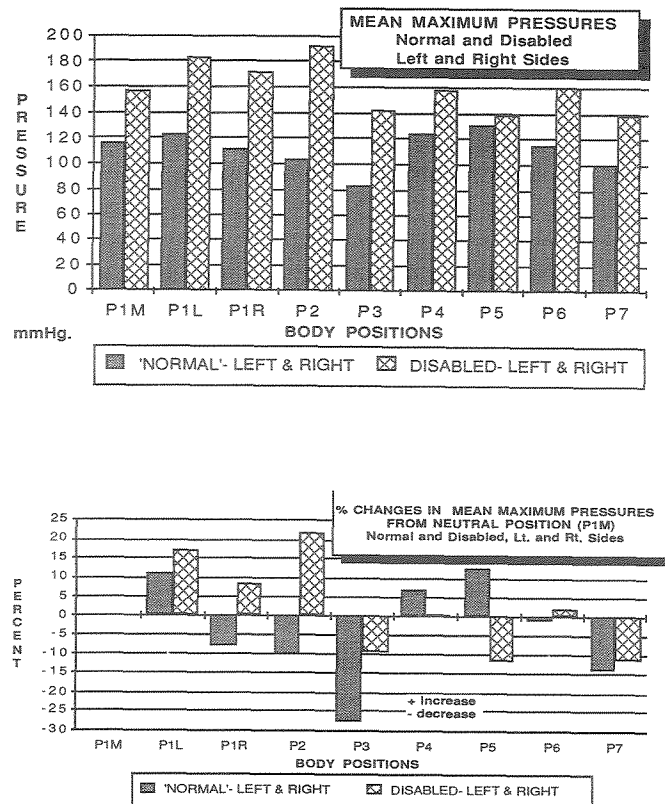


Figure 7. Results of maximum pressure measurements. Upper plot shows maximum pressures for both groups in nine postures. Lower plot shows the percentage of change of maximum values from those recorded in the neutral position (P1M).

2. Mean maximum pressures of individuals with SCI are significantly higher than nondisabled individuals in all nine sitting postures studied. In the neutral posture, on average, the maximum pressure for a person with SCI is 26 percent higher. The differences range from a low of 6 percent in the P5 position to a high of 46 percent in the P2 position.
3. The maximum pressures can be reduced by postural changes from those values occurring in the neutral P1M posture. These reductions are: lateral trunk bending 15° (-32 to -38 percent), forward flexion to 50° (-9 percent), backrest recline to 120° (-12 percent), and full-body tilt to 20° (-11 percent).
4. Alterations to sitting postures cause similar changes to the A/P location of the maximum pressure (P_{mx}) for both disabled and nondisabled. In the neutral posture, forward trunk flexion of 30° and of 50° results, on the

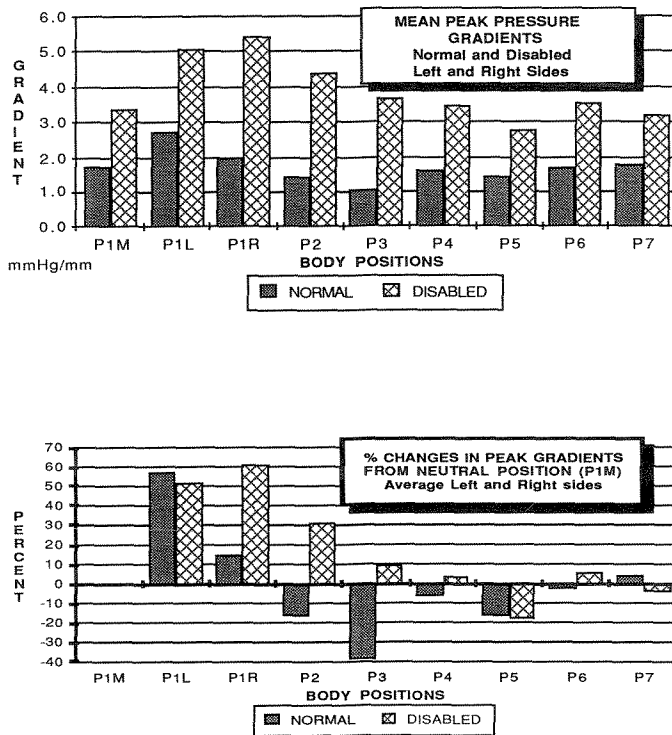


Figure 8.

Computed results of peak pressure gradients. Upper plot shows the peak gradient for both groups in each of the nine postures. Lower chart indicates the percentage of change in the peak gradients from the neutral position value (P1M) that results from changes in seated posture (P1L-P7).

average, in a posterior shift of 2.4 and 2.7 cm, respectively. The largest posterior shift of 6 cm occurs with the backrest reclined to 120° (P5).

- On average, individuals with SCI have peak pressure gradients that are 1.5 to 2.5 greater than nondisabled people (Figure 8). Among the postures studied, maximum reduction of the peak gradients from those measured in the neutral posture occurs after backrest recline to 120° (P5, -8 percent). Large increases in pressure gradients can occur during lateral bending (50-60 percent) or during forward trunk flexion (10-60 percent).

Tangential shear study

Figure 9 graphically depicts the results of the surface shear-force study. The results may be summarized as follows:

- TIS acts at the body-seat surface in all nine postures studied.
- With respect to the neutral posture (P1M), maximum reductions of TIS force occur upon forward trunk flexion of 50° (P3) (-133 percent or a reversal of 33 percent) and upon full-body tilt to 20° (-85 percent). Extrapolation of these results suggests that full-body tilt to approximately 25° reduces the TIS to near zero (-100 percent). Further tilting causes a reversal and a gradual increase in TIS in the opposite direction.
- Compared with the values in the neutral posture, backrest recline postures, P4 and P5, cause an increase in the TIS by 7 percent and 25 percent, respectively.

DISCUSSION AND IMPLICATIONS

The average pressure distribution across the critical buttock area varies little between nondisabled persons and individuals with paralytic injuries. Furthermore, typical alterations in postures assumed while sitting in a wheelchair have minimal effect on the average pressure distribution. Forward flexion of the trunk to 30° is required to obtain a reduction of about 15 percent from the average pressure values in the neutral posture.

People with SCI have maximum pressures that are significantly higher than nondisabled individuals in most postures typically assumed in a wheelchair. The differences in mean maximum pressure range from a minimum of 6 percent in the backrest recline 120° posture to a maximum of 45 percent in the forward flexed posture to 30° (P2). The difference in the neutral posture (P1M) is about 25 percent. It is concluded that these inherent differences in mean maximum pressures are largely due to asymmetrical ischial loading caused by spinal/pelvic deformities and atrophy of tissue over the ischiae.

Changes in posture can effectively reduce the maximum pressures of SCI people from those values that are present in their neutral sitting postures. On average, recline of the backrest to 120° reduces the neutral values by 12 percent as does the full-body tilt to 20° (11 percent). Forward trunk flexion to 50° yields a reduction of 9 percent. Lateral bending to 15° reduces the neutral values by 30-40 percent on the unweighted side.

