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Evaluation of Transducer Performance for Buttock-Cushion Interface Pressure Measurements[a]

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Abstract—To assess the performance of transducers used clinically to measure pressure at the skin-cushion interface of seated patients, transducers were placed between slabs of gel and/or foam materials compressed between platens. The recorded pressures consistently exceeded the nominal pressures calculated using the surface area of the slabs. This overestimation, observed in both miniature diaphragm transducers and air cell transducers, appeared to result from preferential loading of the transducer due to insufficient structural compliance in the environs. On the other hand, air cell transducers placed at a skin-foam interface beneath the thighs of human subjects gave readings which agreed closely with subcutaneous tissue pressure measurements obtained from a wick catheter inserted at the same location. These results suggest that, although pressure measurements are prone to error due to load sharing, results obtained clinically from subjects on soft cushions are reasonably accurate because of the high compliance of human soft tissue and the foam. Under low loads these distribute the pressure equitably and avoid concentrations of load on the transducer.

INTRODUCTION

Pressure measurements at the buttock-cushion interface are used widely in the management of decubitus ulcers in wheelchair-bound patients. In current rehabilitation practice, cushion prescriptions are based largely on interface pressure measurements obtained by means of a wide variety of transducers; this makes the accuracy and reliability of transducer responses essential factors in effective cushion prescription. Unfortunately, these transducers have different response characteristics and the response of a given transducer type may also depend on the type of cushion under test.

In a prior study, this group compared the clinical performance of Kulite electronic transducers with that of pneumatic Scimedics Pressure Evaluator Pads in the course of clinical measurements involving seated subjects (3); that work suggested that some differences in transducer performance in relation to support material may exist. Also, Patterson and Fisher (4) have recently reported experiments designed to evaluate transducer performance at the interface between a pneumatic cuff and skin; they found a wide difference in results between various miniature-diaphragm, straingage-type transducers. Their study did not include pneumatic transducers or assess the effects of different interfacing materials.

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The many types of transducers available for interface pressure measurement include those based on electrical resistance and capacitance, as well as pneumatic "air-cell" and air flow types (1,2). Of these, the semiconductor (based on electrical resistance) and "air-cell" transducer types are in extensive clinical use to measure buttock-cushion interface pressures.

In addition to questions about transducer performance raised by these studies, there is another unknown involved in the measurement of buttock-cushion interface pressure: the relationship of the measured pressures to actual pressures within the tissues. There is clearly a need to compare the interface pressure data with subcutaneous pressure measurements in the loaded tissues. Consequently, the purpose of this study was twofold: first, to evaluate further the relative performance of various interface pressure measurement transducers when used on different seating surfaces, and second, to investigate the relationship between the interface transducer readings and subcutaneous pressures. The first question required bench testing with transducers at different types of interfaces, the second required in vivo tests using human volunteers. Each type of experiment will be described and discussed separately.

TRANSDUCER PERFORMANCE AT AN INTERFACE

Materials and Methods

In the present study, several types of transducers were tested; they are listed in Table I, together with their sources and dimensions. Two of those tested, the miniature single-cell and the Scimedics transducers, are air-cell types. Such a device has electrical contacts bonded to opposing inner surfaces of a small flexible bladder which can be hand-inflated with air. As the bladder is inflated, the internal pneumatic pressure at which the electrical contacts separate is assumed to equal the external pressure on the bladder. These transducers do not provide a continuous measure of the interface pressure.

The Precision and Kulite transducers are semiconductor, piezoresistive, diaphragm-type transducers. The electrical resistance of the sensing material changes as the load applied to the diaphragm changes. The change in resistance is measured via a Wheatstone bridge and requires an excitation voltage. The signal from the bridge can be amplified and recorded continously.

Transducers were calibrated using a deadweight, compressive loading device designed previously by the authors (5). The transducer under

TABLE 1

Transducer model	Dimensions	Source
LQS-125-200 O–200 PSI	O.D., 4 mm Diaphragm diam; 2 mm Thickness: 0.8 mm	Kulite Semiconductor Products, Inc. 1039 Hoyt Ave. Ridgefield, NJ 07657 (201) 945-3000
Model 156 O–25	Length: 9.0 mm Width: 5.0 mm Thickness: 1.0 mm Diaphragm diam: 4 mm	Precision Measurement Co. P.O. Box 7676 Ann Arbor, Michigan 48107 (313) 995-0041
Scimedics Pressure Evaluator Pad	90 mm x 100 mm oval Thickness: 0.5 mm	Scimedics Contemporary Products P.O. Box 4444 Anaheim, CA 92801
Miniature single air cell (experimental design)	Diam. of contact area: 4.00 mm Width of cell: 23 mm Length of cell: 280 mm Thickness: 0.25 mm	Experimental design, not commercially available

test was sandwiched between two square slabs or blocks of soft material. The upper block was chosen to represent human flesh, while typical cushion materials were used as the lower block. Measurements were taken at the interface between layers as would be done between buttock and cushion during routine clinical measurements. Loads were applied on the upper block using a round plate which exceeded the dimensions of the compressed blocks, so that the area of the blocks of material determined the nominal applied stress or pressure. Dimensions of the slabs were 127 mm x 127 mm x 25 mm (5 x 5 x 1 in) for most tests; in certain cases, tests were repeated with 100 x 100 mm (4 x 4 in) blocks.

PVC (polyvinyl chloride) gel was used as an upper slab to represent human soft tissues. The PVC gel is an incompressible hyperelastic material with nonlinear material characteristics. The material characteristics of PVC gel (6) were thought to be a reasonable representation of the incompressible hyperelastic and nonlinear mechanical properties of human soft tissues, for the purpose of the bench tests.

Selected cushion-material types of gel, foam, and a hard surface were used as lower blocks to simulate a seating surface. Under compression, the foam material used in this study had a nonlinear stress-strain relationship. At a strain level of 0.2, the tangent Young's modulus was 11.3 kPa and the Poisson's ratio was 0.15 for the foam. At the same level, the PVC gel had a tangent Young's modulus of 22 kPa, and a Poisson's ratio of 0.50.

To study the comparative performance of the four types of transducers, each device was first calibrated pneumatically. In the case of the electronic transducers, the excitation voltage recommended by the manufacturer was used. After the pneumatic calibration, each transducer was placed at the interface between the two materials, and loads were applied as described. The total applied load was divided by the surface area of the interface between the slabs to calculate the nominal applied compressive stress. Only one transducer was sandwiched between the blocks at any given time. The response of each transducer was noted with applied stresses of 0 to 20.7 kPa (0–156 mmHg) in steps of 3.45 kPa (26 mmHg), for each of the three seating materials—foam, gel, and hard surface. For each test, adequate time was allowed so that measurements were essentially made in static equilibrium.

Five observations were made for each load case. Statistical significance of the differences, between the means for each load case, were determined using t-tests.

Results

With a single exception, all transducers gave significantly high readings when compared to the actual values of nominal (applied) stress calculated as load/block area ($P < 0.005$). The transducer responses (means and standard deviation from 5 repeated measurements, with gel-gel, gel-foam, and gel-hard surface interfaces) are presented in Tables 2, 3, 4. Performances of all transducers were closer to "ideal" (nominal) expected pressure at the gel-foam interface than at the other two types of interfaces. Transducer accuracy was very poor at the gel-gel and gel-hard surface interface conditions, leading to errors approaching 100 percent as shown in the tables.

With respect to the effect of slab size, performance of the Scimedics transducer was improved at the interface between the smaller (100 x 100 mm) blocks, as compared to the larger 127 x 127 mm blocks (Tables 2-4).

Discussion

Cushion performance is often judged, using the interface pressure measurement data with the implicit assumption that interface pressure transducers are accurate and reliable. The results of the present study suggest that the performance of a given transducer is highly dependent on the properties of interface materials and on the ratio of transducer surface area to the contact area of the interfacing materials (contact geometry).

All transducers have a finite thickness, which creates a certain "gap" between the two surfaces. The transducer therefore tends to support the load, causing local concentrations of stress, with the intensity of the concentrated stress depending on the compliance and thickness of the transducer. This problem tends to be minimized if the transducer surface area is either very small, or is equal to the total area of the interfacing surfaces.

For example, when the surface area was 1.61 x $10^{-2}m^2$, (25 in^2), the Scimedics transducer produced a larger error than it did with the smaller block (1.00 x $10^{-2}m^2$), because in the latter case the entire load was transmitted by the transducer which was nearly the same size in support surface. With the larger block, part of the load was supported by the material. Ideally, the transducer should either support all of the load, or should share it uniformly with the support surface, in effect matching its structural impedance. In the former case, the transducer actually becomes a load cell as the transducer surface area approaches the interface contact area.

The enveloping property of the interfacing material (its ability to "wrap around" the trans-

TABLE 2
Transducer responses* at gel-foam interfaces

Applied Nominal Stress kPa	Precision kPa	Kulite kPa	TIRR Single Cell kPa	Scimedics in 127 x 127mm blocks kPa	Scimedics in 100x100mm blocks kPa
3.45	3.60 +/- 0.24	3.7 +/- 0.1	2.17 +/- 0.31	3.55 +/- 0.07	3.55 +/- 0.11
6.89	8.39 +/- 0.34	7.7 +/- 0.24	6.45 +/- 0.57	8.16 +/- 0.15	7.55 +/- 0.09
10.34	13.78 +/- 0.71	12.98 +/- 0.34	11.27 +/- 0.66	13.7 +/- 0.37	11.9 +/- 0.15
13.79	18.9 +/- 0.81	18.21 +/- 0.48	16.33 +/- 1.0	19.65 +/- 0.26	16.57 +/- 0.16
17.24	24.4 +/- 1.16	23.69 +/- 0.6	21.96 +/- 1.52	25.83 +/- 0.11	21.13 +/- 0.12
20.68	29.78 +/- 1.32	29.31 +/- 0.72	27.47 +/- 1.82	30.52 +/- 1.97	25.82 +/- 0.11

*Each of these quantities represents the mean of 5 repeated tests. Standard deviations are shown next to the mean values. Each of these readings is significantly different from the corresponding applied nominal stress ($P < 0.005$).

TABLE 3
Transducer responses* at gel-gel interfaces

Applied Nominal Stress kPa	Precision kPa	Kulite kPa	Miniature Single Cell kPa	Scimedics/ 127 x 127mm blocks kPa	Scimedics/ 100x100mm blocks kPa
3.45	6.48 +/- 0.3	6.82 +/- 6.82	6.34 +/- 0.42	8.14 +/- 0.3	5.53 +/- 0.20
6.89	12.16 +/- 0.35	12.32 +/- 0.84	12.75 +/- 0.39	16.33 +/- 0.37	11.13 +/- 0.21
10.34	17.67 +/- 0.47	17.54 +/- 0.32	18.64 +/- 0.35	23.84 +/- 0.4	16.73 +/- 0.22
13.79	22.68 +/- 0.55	22.59 +/- 0.25	24.37 +/- 0.48	31.58 +/- 0.49	22.46 +/- 0.12
17.24	26.99 +/- 0.83	27.46 +/- 0.34	29.96 +/- 0.27	39.03 +/- 1.62	27.98 +/- 0.13
20.68	31.44 +/- 0.85	31.98 +/- 0.45	35.31 +/- 0.4		33.70 +/- 0.11

*Each of these quantities represents the mean of 5 repeated tests. Standard deviations are shown next to the mean values. Each of these readings is significantly different from the corresponding applied nominal stress ($P > 0.005$).

ducer), is an important factor affecting transducer performance. PVC gel apparently did not envelop the air cell transducer as well during inflation as did foam. Also, the gel apparently enveloped the semiconductor transducers poorly. With the gel-gel interface, the structural impedance mismatch created by transducers between the surfaces is great, causing larger errors. In the case of gel-foam, the foam envelops the transducer, reducing the load-supporting effect of the transducer when compared to other surfaces. (All transducers gave the lowest readings with gel-foam.) As is known, an ideal transducer for measuring interface pressures would have infinitesimal thickness and match the structural properties of the material. Unitl that is achieved, these tests suggest that some correction factor may be necessary when making clinical seating pressure measurements—if more than comparative measures of pressure on the same seating material are required.

Since buttock shape and surface are different from those of the slabs used in tests, it is difficult to directly extrapolate the bench-test results to human cases. In fact, prior clinical measurements (3,5) suggest that material-related variations generated in measurements on actual buttock-cushion

TABLE 4
Transducer responses* at gel-hard interfaces

Applied Nominal Stress kPa	Precision kPa	Kulite kPa	Miniature Single Cell kPa	Scimedics/ 127 x 127mm blocks kPa	Scimedics/ 100x100mm blocks kPa
3.45	5.29 +/- 0.97	6.38 +/- 0.87	5.96 +/- 0.4	6.76 +/- 0.6	5.07 +/- 0.17
6.89	9.74 +/- 1.02	11.98 +/- 1.02	11.69 +/- 0.79	13.10 +/- 0.61	10.08 +/- 0.21
10.34	13.92 +/- 1.14	18.87 +/- 1.18	18.08 +/- 1.93	19.62 +/- 0.95	15.01 +/- 0.32
13.79	18.2 +/- 1.28	24.96 +/- 1.23	24.34 +/- 1.11	26.25 +/- 0.81	19.99 +/- 0.24
17.24	22.83 +/- 1.3	31.59 +/- 1.21	31.50 +/- 1.3	32.35 +/- 0.87	25.08 +/- 0.17
20.68	27.49 +/- 2.12	37.63 +/- 1.16		38.71 +/- 0.87	30.12 +/- 0.26

*Each of these quantities represents the mean of 5 repeated tests. Standard deviations are shown next to the mean values. Each of these readings is significantly different from the corresponding applied nominal stress. ($P > 0.005$).

interfaces exist, but are far lower than those seen in this series of bench tests. The explanation for this important observation probably lies with the special nature of living skin and subcutaneous tissue. Although the PVC gel material simulates human soft tissues to some extent, there are wide differences in certain respects. First, the soft tissues of the buttock consist of three distinctive layers with different material properites (7). Second, soft tissues are made up of anisotropic, viscoelastic, and discontinuous materials containing numerous blood vessels, and other structures such as glands and fascia layers. Third, the stress/strain characteristics of the soft tissues and the gel differ considerably in the lower strain regime (Fig. 1)

In the low-strain regime, the elastic modulus for skin tends to be very low, whereas PVC gel has a more linear response throughout the loading curve. Furthermore, during loading, events such as blood displacement, flow of interstitial fluid, and slip between tissue layers also may occur. As a result of these more favorable mechanical characteristics, soft tissues can envelop an object more completely than the PVC gel, and are thus better at reducing any mismatch in structural impedance and thus at equilizing load support. Since stress

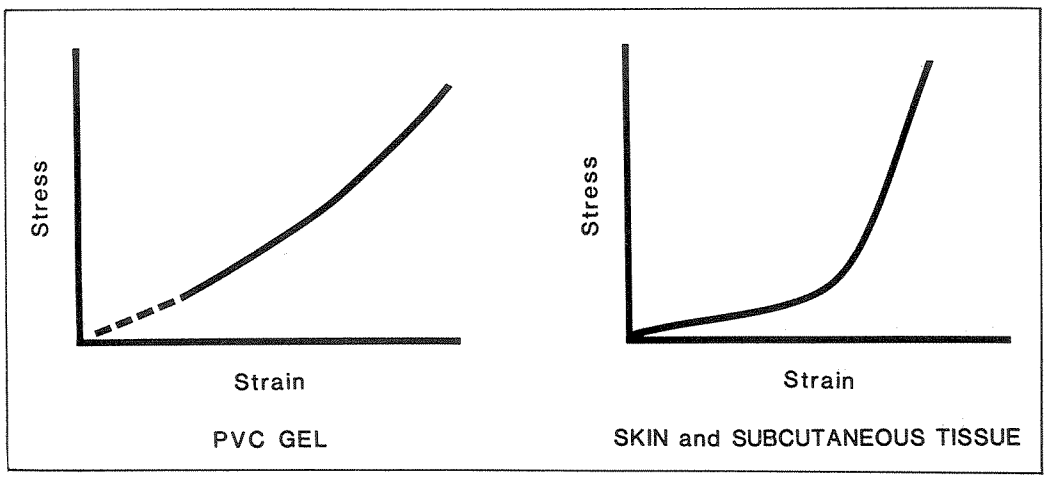

FIGURE 1
Comparison of stress-strain characteristics in compression of PVC gel versus skin and subcutaneous tissue. Note the tendency toward greater compliance in the low-strain area for the soft tissues. PVC gel has a more linear response.

concentrations decrease with increasing envelopment of the transducer, transducer performance tends to improve when in contact with actual flesh. Nevertheless, our data indicate that currently used pressure transducers have a tendency to cause overestimation of the interface pressure, and that responses do vary with the cushion material. Further work is needed to document appropriate correction factors for clinical measurements on different materials (the next portion of this paper reports a first step in that direction). Caution is also required in interpreting measurements from individuals with lean or atrophied buttocks, or measurements in other body areas where the enveloping qualities are reduced, as between a limb stump and a prosthesis socket.

COMPARISON OF INTERFACE PRESSURES WITH SUBCUTANEOUS PRESSURES

Materials and Methods

Wick catheters are widely used for measuring subcutaneous interstitial fluid pressure in vivo (8). The procedure, as described in detail by Snashall et al (9), involves placing fluid-filled wick catheters in the subcutaneous tissues using thin-walled needles; the catheters are prepared by pulling dermalon fibers into one end of 0.58-mm I.D. polyethylene tubing.

In this study, wick catheters were employed to investigate the relationship between pressure at a skin-foam interface (as measured by a Scimedics Pressure Evaluator Pad) and the subcutaneous tissue pressure.

To perform these tests, wick catheters were placed bilaterally in the posterior thighs of human subjects. The thigh was selected as an experimental site because the underlying bone structure of the femur permitted more accurate locating of the wick in relation to bone than could be achieved reliably over the ischial tuberosities. (In the buttocks, the local position of the wick catheter in relation to the ischium would be difficult to determine). The catheters, the associated physiological pressure-measurement transducer, and the calibration system were all sterilized; then the transducers were calibrated in a sterile system immediately prior to insertion. Each wick catheter was filled with heparinized saline and connected to a separate physiological transducer (Alitech M52OE) by means of a fluid bridge. The transducer signals were amplified and recorded by means of a Gould Transducer Amplifier (Model 13–4615–50) and Gould-Brush Recorder (Model 260). For calibration purposes, the wick catheter was inserted through a rubber stopper into an extension tube filled with saline and connected to a manometer.

With the subject in a prone position, local anesthesia was induced with one percent Xylocaine in the thighs bilaterally at the point of catheter entry. The pressure readings were adjusted to zero with transducers and wicks positioned at the level of the insertion site. The wick catheters were then placed bilaterally in the thighs through thin-walled 18-gauge needles. Care was taken to see that the wicks were close beneath the dermis and at least 30–50 mm away from the point of entry into the skin. The wick location could easily be identified, and could be tested for response by applying slight finger pressure to the skin, while monitoring transducer readings. After adjustment of the wick catheters, deflated Scimedics transducers were positioned on the skin over the wicks bilaterally and anchored minimally with tape. The subject was then transferred to a special seat. Small quantities of heparinized fluid (0.1–0.2 ml) were injected into the tissues through the wick catheters at regular time intervals to retard the possibility of clotting, and to insure measurement of total tissue-pressure. The experimental setup is shown in Figure 1.

Single, sequential readings of interface and wick pressures were obtained bilaterally in three subjects with feet hanging free and with added loads of 67 and 111 Newtons (15 and 25 lbs.-f) placed on the feet in order to produce nominal pressures beneath the thigh which approached those normally occurring in the human buttocks while seated on a cushion. The air-cell transducer was always deflated (thickness 0.5 mm) at the time of wick catheter pressure measurements. When one foot was being loaded, the other was kept hanging free. On each subject, the procedure was repeated with pads of three different thicknesses to generate three different pressure ranges for each of the three test loads; the subject was allowed to stand between sets of measurements on each pad.

Results

Table 5 shows the relationship between mean subcutaneous pressures as measured by the wick catheter and the mean interface pressures as measured by a Scimedics Pressure Evaluator Pad, under three loading conditions from three human subjects. Pressures ranged from 3.98 kPa with no load on the foot to 9.97 kPa with a 111.2-N load on the foot. The interface and subcutaneous pressure measurements in the thighs correlated well. The differences between subcutaneous and inter-

TABLE 5
Subcutaneous and interface pressure measurements * beneath loaded thigh.

	No Load on Foot		Load of 66.72 N on the Foot		Load of 111.2 N on the Foot	
Cushion Type	Subcutaneous Pressure kPa	Interface Pressure kPa	Subcutaneous Pressure kPa	Interface Pressure kPa	Subcutaneous Pressure kPa	Interface Pressure kPa
2" Foam (Rogers #3040)	4.44	4.60	5.60	5.21	6.54	6.24
1/2" Foam (Rogers #1836)	4.17	3.98	6.04	5.69	9.92	8.28
1" Soft Open Cell Foam	4.20	4.80	7.83	7.10	9.97	8.13

* Subcutaneous pressure was measured with a wick catheter. Thigh-cushion interface pressure was measured with "air-cell" (Scimedics Corp.) transducer. These values represent the mean of readings taken from three human subjects. The differences between interface and subcutaneous pressures were statistically insignificant ($P > 0.75$). Thin foams were employed as cushion materials so as to generate interface pressures comparable to the clinically observed buttock-cushion interface pressures. (1 kPa=7.6 mmHg)

face readings were obviously small and proved to be statistically insignificant as revealed by paired and unpaired two-tailed t-tests ($P > 0.25$).

Discussion

The wick catheter technique is a reliable method of measuring interstitial fluid pressure. The wick permeates a relatively large volume of interstitial fluids and permits measurement of pressure developed in these fluids. This technique has been used by a number of investigators for measuring interstitial fluid pressures in animals and in humans (9,10).

In our laboratory, the reliability of the wick method was studied further in animal experiments using pigs. It was found that approximately 70 percent of pressure applied by an external circumferential pressure cuff was transmitted to the interstitial fluids in normally-hydrated states. However, if the microenvironment of the wick was altered by injecting very small volumes of saline (0.02 ml), then nearly 100 percent of the externally applied pressure appeared to be transmitted to the fluid (11, 12). Thus, in these experiments with human subjects, the fluid pressure measured by the wick equaled the total pressure in the tissue—a combination of interstitial fluid pressure and externally applied pressure. Care was taken to place the catheter close to the surface beneath the dermis to minimize effects of stress distribution by the tissues.

The loads hung on the feet were selected to

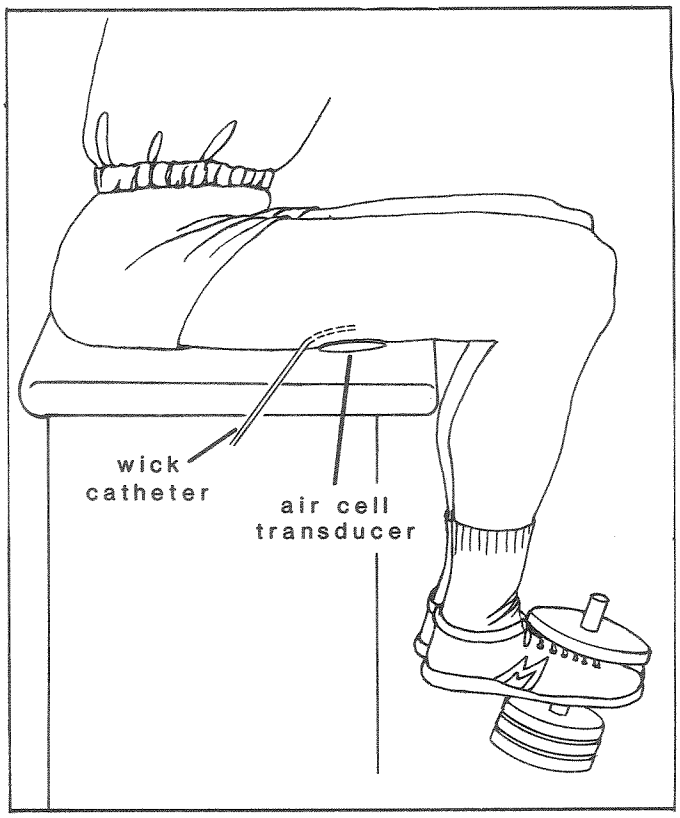

FIGURE 2
The experimental setup for comparison of interface and interstitial pressure measurements. Indentation of the soft tissue by the air cell transducer was minimal but is exaggerated here to show the location. Muscles were relaxed during measurements so dead weight loading on legs compressed thigh against table surface covered by a 25-mm-thick soft foam pad.

generate pressures underneath the thighs in the same range as those registered beneath the buttocks during sitting. The close correlation between the subcutaneous pressure measured with the wick catheter, and the pressure at the skin-pad interface as measured by the Scimedics Pressure Evaluator Pad, suggests that these devices can be regarded as valid clinical measurement tools. Even so, it must be recognized that these transducers reflect average rather than peak pressures within their measurement area (13), and that gels or other relatively stiff seating materials may tend to distort results.

SUMMARY

Two types of semiconductors/transducers and two types of pneumatic transducers were evaluated in vitro for use in clinical measurement of skin-cushion interface pressures. During bench testing in a two-layer system, all transducers gave readings considerably higher than the calculated nominal (applied) stress. The accuracy of transducer responses was dependent clearly on the properties of the interfacing materials and the relative sizes of the pads and transducers. In contrast, the readings from an air-cell type transducer beneath the thigh in human subjects appeared accurate in that it correlated well with subcutaneous interstitial fluid pressure as measured with a wick catheter.

This work draws further attention to the **potential** inaccuracies that can result from employment of small, thin transducers to measure pressures at the interface between materials. On the other hand, the study suggests that, with care, reasonable results can be obtained clinically owing to the favorable compliance of human tissue under low strains: the tissue acts to distribute the load evenly over the transducer.

While these tests with normal human subjects on one type of seating material were favorable, the bench tests drew attention to the system's high sensitivity to the mechanical properties and configurations of the interfacing materials. For that reason, interpretation of results of clinical pressure measurements continues to require considerable caution, particularly when comparing cushions of varying stiffness, when buttock tissues are thin or atrophied, and when clothing or other material intervenes at the interface ■

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