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A Pressure sensitive Mat for Measuring Contact Pressure Distributions of Patients Lying on Hospital Beds

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ABSTRACT

The authors describe a novel system for sensing and displaying the distribution of contact pressure caused by a patient's lying on a hospital bed. The system includes a flexible, pressure sensitive mat, electronics to activate the mat, a small computer to process data, and a color video display. The present prototypes can sense pressure at 1,536 discrete locations in a rectangular grid of 24 x 64 nodes, each node representing an area of 4 cm². The computer receives data from each node and displays the results as a false-color map, refreshable every 5 seconds. The pressure sensitive mat itself includes two orthogonal arrays of ribbon-like conductors, composed of silver coated nylon fabric, which are separated by insulating open cell foam rubber. The system monitors the electrical capacitance between selected pairs of horizontal and vertical conductors on opposite sides of the foam. The crossing points form pressure sensitive nodes. Increased contact pressure compresses the foam, thereby decreasing the distance between the conductors and increasing the capacitance. Node capacitance is determined by measuring the current through it from a voltage source. The outputs of the various nodes are scanned, normalized, and converted to pressures using the known compressive stress-strain relationship for the foam, and the data are then displayed as a false-color image of the pressure distribution.

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This report describes the creation of a contact-pressure mapping system configured as shown in Figure 1. The system was developed as a tool to study the clinical problem of pressure sores (bedsores) and to aid in the design of improved hospital beds. Major components of the system include a pressure sensitive mat that can be placed on top of a mattress to sense the weight distribution of a person lying on it, an interface unit for supplying electrical driving signals to the mat and for receiving pressure-related signals from the mat, and a small computer (IBM/XT) with a floating-point processor and color graphics capability. The system creates images of the distribution of contact pressures caused by a patient lying on a hospital bed with a spatial resolution of about 2.5 cm. Such images allow the user to characterize high pressure regions that contribute to the genesis of pressure sores. In addition to providing research data, the system may serve as the progenitor of a family of computer controlled hospital beds that have the ability to identify high pressure points and then to automatically contour the underlying surface to eliminate them.

Figure 1. Overview of prototype system for contact pressure mapping. (a) The current version includes a pressure sensitive pad incorporating 1,536 discrete pressure sensitive nodes, an analog-to-digital interface unit, and a PC-based video display, on which images of the pressure distribution are presented. (b) Detail of video output as originally conceived. This image is a contour interval plot created by off-line processing of uncalibrated data from an early prototype. Later versions of the system utilized a false-color, pixel-by-pixel display that was computationally much faster to create.
MATERIALS AND METHODS

Principle of Transduction

The basic construction of the pressure sensitive mat is illustrated in Figure 2. It includes a compressible dielectric layer with two orthogonal arrays of flexible conductors, disposed on opposite sides. The dielectric layer is an open cell foam having an unloaded thickness of about 5 mm. The flexible conductors are made of silver coated nylon fabric. The pressure sensitive mat thus constitutes an array of soft, variable capacitors.

Figure 2. Simplified sketch of pressure sensitive mat construction. An array of variable capacitors is formed by orthogonal strips of conducting fabric, separated by compressible, insulating foam material. Crossing points of x and y elements constitute pressure sensitive nodes. As normal force is applied to a node, the foam is compressed, and the node capacitance increases.
Pressure sensing is accomplished by measurements of the capacitance between selected pairs of orthogonal conductors. This capacitance is confined largely to the zone of intersection and is a function of contact pressure. Each such crossing point is defined as a pressure sensitive node. The variation in capacitance caused by compression of the foam constitutes the principle of transduction: as the separation of conductors is reduced by increasing normal force, the capacitance of a node is increased. The grid-like arrangement of orthogonal rows and columns leads naturally to computer controlled sampling of pressures at various x, y coordinates on the surface and to generation of a matrix type display.

The equivalent circuit that is formed by the structure of Figure 2 is illustrated in Figure 3. By way of example, a selected node, C₀, and its eight nearest neighbors, C₁ through C₈, are illustrated. One terminal of each capacitor in a row of the matrix is connected to one of the terminals of the demultiplexer, and the other terminal of each capacitor in a column of the matrix is connected to one of the terminals of the multiplexer. When it is desired to measure the capacitance of a particular node, a driving signal having a known sinusoidal voltage is applied to the row in which the node is located. The current from the measured capacitance is sensed via the associated column electrode, which is connected to a current-sensing amplifier with low (ideally zero) input impedance. The capacitance at the node is determined by the thickness of the dielectric layer, which in turn is a function of pressure. A higher capacitance represents a thinner dielectric layer and hence a greater applied pressure.
Two sets of digitally controlled analog switches are used to sample signals from the array of
nodes. The first, connected to a 5-kHz, 10-volt, sinusoidal voltage source, functions as a
demultiplexer. The second, connected to the sense amplifier, functions as a multiplexer. The
particular generator outputs of 5 kHz and 10 volts are not critical to system performance, and
represent compromise values based on considerations of patient safety, interference with other
equipment in the environment, sensitivity, speed of sampling, and ease of circuit fabrication.
Under computer control, these switches are used to measure the current through each node in
rapid succession. When it is desired to interrogate a node defined by a particular x, y coordinate,
the sense amplifier is switched to the x-th column of conducting fabric, and the voltage source is
switched to the y-th row. In practice, a stable signal is obtained about 1 msec after both
connections are made. After sampling, the next row or column in the array is then interrogated.

Elimination of Inter-node Interaction

One factor that must be addressed in the design of such a system is the potential electronic
interaction among nodes. In general, the current that is sensed in one column electrode is not
determined exclusively by the capacitance at the crossing point. Although the primary path from
the driving electrode to the sensing electrode is through the capacitor $C_0$ in Figure 3, the driving
signal can also be coupled to the sense line via unselected capacitors $C_3$, $C_2$, and $C_1$. If the
unselected rows are allowed to float electrically, some signal can be capacitively coupled in this
way form the drive line to adjacent columns, then to adjacent rows, and ultimately back to the
sense line. Changes in pressure (separation) at adjacent nodes can alter the amount of capacitive
coupling, and so the measured signal in the sense line.

In the present system, the effects of the surrounding nodes on the measured signal are eliminated
by connecting all of the unselected electrodes, other than the ones associated with the row and
column of interest, to ground. In practice, single-pole-double-throw switches connect the row
electrodes either to the voltage source or to ground. Similar switches connect the column
electrodes either to the current-sensing amplifier or to ground. Thus, all unselected electrodes in
the entire array are clamped at ground potential. As a consequence, spurious current coupling
between drive and sense electrodes through chains of unselected nodes is blocked, because the
voltage across at least two nodes in each such chain is zero. This design feature ensures that
changes in the capacitances of unselected nodes introduce no error in the output signal. The fact
that the column electrodes are clamped at ground potential and that the voltage on the selected
column electrode is, ideally, zero also eliminates error that would otherwise be caused by the
stray capacitance of the cable connecting the interface to the pressure sensitive mat.

Mat Design

During the course of development, pressure sensitive surfaces containing 2 x 2, then 16 x 16, and
finally 24 x 64 member arrays of pressure sensitive nodes were fabricated. The current method of
construction for the mat is illustrated in Figure 4. It comprises a symmetric integration of two
mats of the general type depicted in Figure 2. A central sensing electrode layer has insulating,
open cell foam layers disposed on both sides, which are in turn encompassed by driving
electrode layers. The three-electrode layers are composed of supporting non-conductive fabric, to which the linear electrodes, composed of conductive fabric, are attached.

The electrically conducting fabric was silver coated nylon, obtained from Swift Textile Metalizing Corporation, Hartford, CT 06101. In the current prototypes, conducting strips 2 cm wide are sewn to three layers of supporting cotton fabric, separated by two layers of compressible foam rubber (typically 5 mm thick) to form the five-layer sandwich design in Figure 4. The two capacitors formed at each node are electrically in parallel but mechanically in series. This design effectively doubles the capacitance of each node. The five-layer design also provides shielding and symmetry to the system. Various types and thicknesses of foam can be introduced between the conducting layers via side pockets that can be reversibly closed with Velcro® hook-and-loop tape. Alternate strips of conducting nylon are sewn on opposite sides of the support fabric (dashed lines) to form an over-and-under pattern. This arrangement keeps adjacent strips of conducting nylon from being shorted due to small wrinkles in the fabric of the electrode layers. The nodes are aligned on 2.8-cm centers. There are 24 exterior, driven lines and 64 orthogonal, inner sense lines, making 1,536 pressure sensitive nodes.

![Figure 4. Realistic sketch of the five-layer construction of the most advanced prototype pressure sensitive mat. Three conducting layers are separated by two compressible foam insulators. Parallel sense lines are included in the middle conducting layer; drive lines are included in the top and bottom conducting layers. Pressure sensitive nodes include two capacitances mechanically in series but electrically in parallel. Over-and-under sewing of conducting fabric on opposite sides of cotton backing material (dashed lines) minimizes shorting of silver coated nylon thread from adjacent conductors in the same layer. The five-layer design provides symmetry and shielding, and also doubles the electrical signal.](image-url)
The two exterior rows of conductors are electrically connected and driven by the 5-kHz voltage source, and the columns of conductors in the middle layer are used for sensing. Accordingly, each measuring node is defined by the area of intersection of a sensing electrode with an aligned pair of driving electrodes. If the two electrodes in the upper and lower driving layers are not perfectly aligned, the measured signal is not adversely affected; rather, any misalignment tends to degrade the spatial resolution of the node.

In operation, the computer sends control signals to the interface unit to sequentially scan each of the pressure sensitive nodes. This scanning is carried out through the coordinated addressing of the driven electrodes and the sensing electrodes. The output signal from an individual pressure sensitive node is a current. The input signal to the A/D converter of the computer is a voltage. The analog interface circuitry converts the sensed current into a voltage, appropriately scaled for the A/D converter. The computer calculates the applied pressure at each measured node from the sensed signals in accordance with a previously determined calibration function that depends upon the compressive stress-strain relationship for the particular foam inserted to form layers 2 and 4 of the five-layer pressure sensitive mat.

**Foam Characteristics**

Critical to performance of the system are the compressive stress-strain characteristics of the foam rubber layers. The general nature of such curves is presented in Figure 5 for a typical synthetic polyurethane foam. Digitized output voltage is plotted as a function of applied pressure, as a test node is compressed by a flat rubber bladder connected to a mercury manometer. As increasing pressure is applied from 0 to 100 mmHg, the output signal rises in a nonlinear fashion, reflecting increased capacitance of the node. Curves for different nodes, indicated by the different symbols in Figure 5, are not identical but have the same general shape. Various curve fitting routines can be applied to represent the loading curves for the various nodes and in turn compute contact pressure from the digitized output voltage. The present implementation uses a five coefficient polynomial, similar to the functions described by Throne and Progelhof [1] and by Fabishak [2] to represent the calibration curve for each node. By utilizing characteristic coefficients for each individual node, the computer can minimize inter-node variations in the calibration function.
Hysteresis

In the present pressure sensitive mat design, we have exclusively incorporated open cell foams rather than closed cell foams, because closed cell foams proved to be insufficiently compliant to produce readily measurable compressive strain in response to 10 to 100 mmHg stress, the desired working range of the system. A major difficulty encountered in early tests utilizing open cell foams was caused by hysteresis in the compression stress-strain relationship for such materials. The curve traced during unloading of the material was markedly different from that traced during loading (Figure 6), especially that traced during initial loading. This marked hysteresis of some foams, which is well known [3], especially for foams that are not pre-stressed, made inference of pressure from digitized voltage output highly dependent upon the compression history of the foam. Fortunately, natural latex foams (¼ inch slab foam or ¼ inch pin-core foam, obtained from Latex Foam Products, Inc., P.O. Box 525, Ansonia, CT 06401) exhibit substantially less hysteresis than synthetic polyurethane foams (Figure 7). Moreover, the hysteresis of natural latex can be further reduced by pre-compressing the material to 100 mmHg a single time prior to making quantitative measurements. The hysteresis-reducing effects of prior compression last for several hours, permitting repeated measurements with a precision of 2–3 mmHg. In addition, natural latex exhibits negligible creep (i.e., continued slow deformation under pressure) and acquires no permanent compression set under prolonged loading [4]. Accordingly, we have used natural latex foams in all recent configurations of the pressure sensitive mat.
Figure 6. Stress-strain evaluation of high-hysteresis foam (Stephen and Lawyer Type SL65-white, ¼ in). Triangles represent initial compression (loading) curve; circles, unloading curve. Displacement on the vertical axis represents compression strain. Separation of the curves represents an extreme case of foam compression hysteresis.

Figure 7. Hysteresis curve for low-hysteresis, natural latex foam (Latex Foam Products, 2 x ⅛ in bi-layer). Dashed line represents initial compression (loading) curve; solid line, initial unloading curve, which was also nearly identical to subsequent loading curves. Pre-stressing this foam abolished measurable hysteresis, which was evident only on the first compression cycle. Digitized voltage output on the vertical axis is a nearly linear function of the compression stress.
Figure 8. Calculated vs. actual pressures for 16 nodes after software assisted semi-automated calibration. Calculated pressures were derived from 4th-order polynomials, which characterized digitized output voltages, $E_{out}$, for each node as a function of known calibration pressure. Data points represent results of a recalibration experiment. All 16 separate points are plotted and are essentially overlapping, demonstrating good agreement with each other and with actual measured pressures.

Calibration

Figure 8 illustrates calibration data from 16 separate nodes. The data were obtained by application of a known pressure from a large bladder, followed by calculation of node-specific coefficients, and subsequent retesting by application of known pressures. Plotted are calculated vs. actual pressures for the 16 nodes in the post-calibration tests. Calculated pressures fall upon the line of identity to within 2 mmHg. To speed the calibration process for the full sized 24 x 64-node mat, a large 50 x 50-cm$^2$ bladder was fabricated and inflated against successive areas of the mat while supported in a wooden frame. The total force on the frame at 100 mmHg, inflation was 748 lb. It was this value of total force that limited the size of the calibration frame to considerably less than the size of the 24 x 64-node mat. However, with software support, the calibration procedure is speeded considerably. Computer assisted calibration software receives signals from a solid state pressure transducer connected to the calibration bladder. With this approach 16 x 16 node regions of the mat can be calibrated simultaneously, as voltage and pressure are automatically recorded. The calibration routine then automatically calculates and stores the coefficients of the node-specific calibration functions, keeping track of the nodes that have and have not yet been calibrated. With this automated system the calibration procedure can be accomplished in about 30 min. The calibration values remain stable for several days, or until a major change in ambient relative humidity occurs, necessitating recalibration.
Note: Humidity, and in turn the moisture content of the foam, was found to be the major factor influencing the stability of system calibration. As foam absorbs moisture, the net dielectric constant of the combined rubber-air-water mixture within the foam changes, owing to the high dielectric constant of water (80 for water, compared with 1.0 for vacuum or air and 2 to 6 for most rubbers). Changes in the net dielectric constant affect the capacitance of the nodes. We found that the capacitance increases roughly 4% for every 1% change in relative humidity. Best results are currently obtained in a humidity-controlled indoor environment Future applications may require addition of a moisture barrier to the pressure sensitive mat.

RESULTS

Figures 9A and 9B are black-and-white renditions of pressure maps created by a supine subject on a hard vs. a soft foam rubber cushioned surface. The pressure sensitive mat was placed between the subject and the test surfaces to create these images. High pressure zones beneath the head, shoulder blades, and sacral areas are clearly visible and are greatly attenuated by the interposition of 4-inch foam rubber between the hard surface and the pressure sensitive mat.

Similar effects are demonstrated in Figure 10 for the same subject in lateral recumbency. Approximately 3 msec are required to scan each of the 1,536 nodes, calculate pressure, and display the result as a colored pixel on the video display. Thus, an entire image can be refreshed about every 5 seconds. When a desired image is obtained, it can be "frozen" on the screen and saved to disk for later analysis. Stored images can be recalled and, if desired, displayed with a different false-color scale.
Figure 9. A, Image of pressure distribution created by a supine adult female subject on a hard wooden surface. The pressure sensitive mat was interposed between the subject and a wooden table. High pressure areas associated with the head, shoulder blades, sacrum, calves, and heels are evident (left to right). B, The same subject on a soft surface. The pressure sensitive mat was interposed between the subject and a 4-inch-thick foam pad. Attenuation of high pressure points is evident, together with spreading of pressure over a larger area. The black-and-white rendition was created by assigning gray-scale values to pressure ranges that are normally represented by colors in the on-line video display.
Figure 10. Same subject as in Figure 10 in lateral recumbency. A, Hard wooden surface; B, Soft foam surface.
DISCUSSION

The present system allows sampling and display of the entire contact pressure distribution created by a patient lying on a hospital bed, at intervals as frequent as every 5 sec. Because the whole patient is imaged, changes in high pressure zones caused by weight shifts can be detected and appreciated within seconds. This feature is important in the evaluation and testing of sleep surfaces, because subtle and unconscious movements of subjects can greatly confound results when single pressure sensors or smaller area arrays are placed under selected body regions.

Moreover, the ability to adapt to changing body position makes such a system potentially useful for future automated hospital beds that employ servo controls and active feedback to minimize high pressure zones. Other potential clinical applications of the system are related to prevention and treatment of pressure sores in bedridden or paralyzed patients [5, 6]. Several colleagues have suggested the potential value of such a device, not only for chronically bedridden or paralyzed patients, but also in anesthesia during prolonged and difficult surgical procedures, when pressure sores are prone to develop. In the latter application the system could continually monitor the patient, identify areas with prolonged, excessive pressure, and alert the surgical team to the need to reorient the potentially endangered area.

The concept of using a pressure sensitive matrix to measure the distribution of contact pressures created by the human body is not entirely new. In the early 1960s, Aronovitz, Reswick, and their colleagues at Case Institute of Technology developed a pneumatic cell matrix to measure the distribution of contact pressures generated by a seated human subject [7]. Pressure was sensed by a multi-cellular inflatable mat utilizing a large number of pneumatically interconnected, flexible cells. Each air cell had an electrical contact on the roof and another on the floor. If local pressure in the mat was greater than the exterior pressure, the cells would inflate, causing the contacts to separate. During data collection the mat was placed under the subject and the pressure in the interconnected cells was gradually raised. An array of 1,886 neon bulbs arranged in a 1-cm grid, isomorphic with the cells of the mat, provided a display related to the spatial integral of the contact pressure distribution in the plane of the mat. With suitable analysis of the pattern of lighted bulbs at different inflation pressures, a contour map of the contact pressure distribution could be obtained. Subsequent to creation of the Case prototype, this same design concept has been embodied in a commercial unit: the Texas Interface Pressure Evaluator (Knightsbridge Medical, Inc., Spring, TX 77379).

Several of the design concepts we incorporated in the presently reported system have been previously described in the patent literature. In 1979, Nicol and Hennig [8] described a pressure sensitive mat constructed from an array of pressure sensitive capacitors. They described two arrays of metallic conductors, oriented at right angles to one another, which formed the plates of the capacitors. The insulator of the capacitors was a compressible dielectric material. The mat was scanned by connecting an alternating current source to one member of one array via a demultiplexer, and the voltage impressed on a selected member of the other array was sensed via a multiplexer. Compared with a pneumatic system, this electrical approach reduces the number of connecting wires, gives the contact pressure distribution directly, and has an inherently faster
response time. The measured pressure dependent variable was voltage, and hence, errors were introduced by stray capacitances. In a later patent, the metallic conductors were replaced by fabric woven with conducting threads. In 1985, Boie and Miller described a similar matrix addressing scheme to measure the pressure-dependent voltage in a 6 x 6 array of force-sensitive capacitors placed in the feedback path of the sense amplifier.

The technical approach that we have followed is also based upon capacitive transduction at the intersections of orthogonal conductive strips. From our experience, however, it is clear that a variety of technical problems remain to be solved before pressure sensitive mats based upon capacitive transduction can be promoted from the status of laboratory research tools to that of clinically useful devices. For example, in applications for which absolute rather than relative pressure data are required, the moisture sensitivity of the calibration must be dealt with: either by including a moisture barrier or by substituting a non-moisture-absorbing foam. Fabrication of continuous sheets of nylon with alternating bands of conducting and non-conducting material would greatly simplify mat construction and provide economies of scale in manufacture. Use of custom made foams specifically designed to be both low in hysteresis and high in moisture resistance would, of course, improve system performance and would be cost-effective on a commercial scale. Future embodiments would also be likely to include a vapor barrier fabric covering for the mat.

The presently existing pressure sensitive mat at our institution has been created strictly as a research tool, not a marketable product. Specifically, the authors have not incorporated all the desirable patient safety features that would be needed for widespread clinical use. The creation of three full scale prototype systems to date, however, does demonstrate the concept and feasibility of contact pressure mapping by differential capacitance sensing under computer control. In addition to research uses in the design of better and more comfortable beds, chairs, and wheelchairs, such devices may find a role as components of feedback-controlled hospital equipment for use by paralyzed or chronically bedridden patients, as training aids for nursing personnel, and as monitors during surgery.

REFERENCES


