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Skin blood flow changes and tissue deformations produced by cylindrical indentors

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Abstract—Since skin blood flow as a function of applied pressure depends on many factors, one may find different curves within each group of subjects, making separation of groups difficult. A dimensional analysis has been out which indicates that the percent decrease in skin blood flow associated with external loading is primarily a function of only three variables: the ratios of bone depth and bone diameter to indentor doameter, and percent compression of the tissue overlying the bone. The load itself is found to be unimportant. It is concluded that measurement of bone depth, bone diameter, and tissue deformation are more important than pressure measurements. Measurements of skin displacement and average indentor pressure for four male subjects indicate that tissue stiffness increases with age, regardless of disability, so that higher pressures are required to produce the same displacement in older subjects.

INTRODUCTION

During our seating studies, it became apparent that we needed to monitor some indicator of tissue response to loading, such as skin blood flow or tissue deformation. However, our attempts to study the results of others investigating skin blood flow were frustrated by the manner in which their data were obtained and presented. Specifically, several different measurement techniques have been employed (1, 2, 4, 5), and skin blood flow has usually been plotted against applied pressure, which causes the results to be different for each individual subject. The use of dimensional quantities such as pressure and flow means that the results must depend on the subject's weight, size, etc. Furthermore, since interface seating pressures are difficult to measure accurately, one often resorts to using an indentor for the application of loading. This means that the results must also depend on the size and shape of the indentor. Also, since the distribution of loading under the indentor is not uniform, one must either measure that distribution or make some simplifying assumptions.

We attempt in this paper to deal with each of these problems by first performing a dimensional analysis of skin blood flow as affected by indentor loading and then addressing the load distribution associated with specific indentor shapes. Associated skin displacements are also considered, and displacement measurements are presented for four male subjects.

DIMENSIONAL ANALYSIS

To ensure measurement of all pertinent parameters and proper presentation of results in their most general form, we first perform a dimensional analysis of the problem, which is represented schematically in Figure 1.

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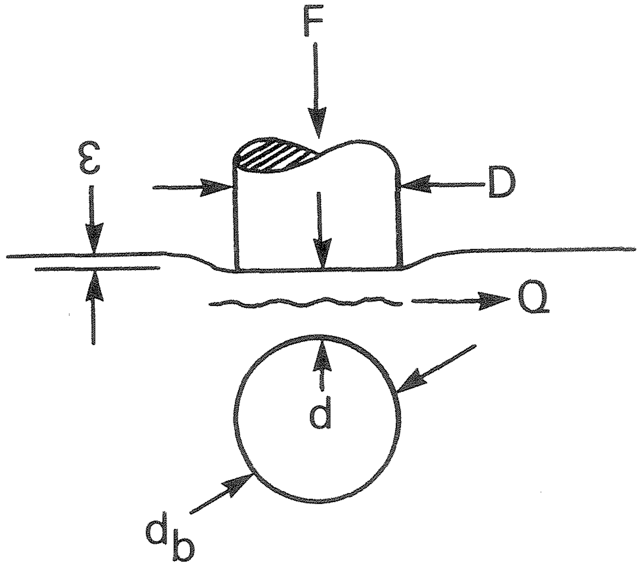

FIGURE 1
Schematic view of skin blood flow under indentor loading. D = diameter; F = applied load; ϵ = deflection of skin surface; Q = volume rate of skin blood flow; d = depth of underlying bone; d_b = diameter of underlying bone.

Here, D represents the diameter of the indentor for application of loads, d is the depth of an underlying bone parallel to the skin surface, ϵ is the deflection of the skin surface caused by the indentor under the applied load F, Q is the volume rate of skin blood flow in the area detected by the blood flow–measuring instrument, and d_b is the diameter of the underlying bone. The subscript zero refers to baseline conditions with no load. We have assumed that the flow reaches an equilibrium value for any given load and therefore is not a function of time. Thus, we can express Q as un unknown function of the above variables, so that

$$Q = f\,(F, Q_0,\ d,\ \epsilon, d_b,\ D) \qquad [1]$$

However, according to dimensional analysis (3), a physical equation makes sense only if the dimensions on the two sides of the equation are the same. To ensure that this is the case, we must first represent each of the above variables in terms of its dimensions, e.g., mass, length, and time. Thus, we replace D by L (length), F by MLT^{-2} (mass × acceleration), etc. By assuming that Q can be expressed as a product of powers of the other variables indicated in Equation 1, we find, upon equating the power of each of the dimensions M, L, and T on either side of the equation, that the condition can be satisfied if the flow parameter Q/Q_0 is a function of the ratios d/D, d_b/D, and ϵ/d. Thus, the relationship can be written in dimensionless form as

$$Q/Q_0 = f(d/D, d_b/D, \epsilon/d) \qquad [2]$$

It is important to note that the force F has dropped out of Equation 2 and that ϵ accounts for the combined effects of applied force and elastic modulus. In fact, since the elastic behavior of the tissue is most likely nonlinear, using actual displacement is preferable to selecting a single (constant) elastic modulus. Finally, it is found experimentally that for transverse loading of the proximal femur in side-lying patients d_b is essentially constant for all subjects. Therefore the ratio d_b/D is essentially constant, so that Q/Q_0 is found to depend on only two nondimensional parameters:

$$Q/Q_0 = f(\epsilon/d, d/D) \qquad [3]$$

This relationship can be displayed in a three-dimensional plot as shown in Figure 2, with each group of subjects represented by a surface in space and defined by the experimental data.

Thus, there are three essential quantities to be measured for a proper experiment in the effect of loading on skin blood flow: the blood flow itself

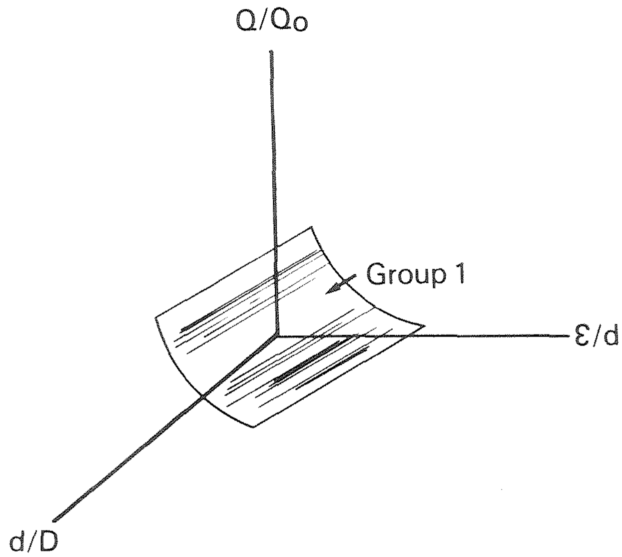

FIGURE 2
Dimensionless presentation of skin blood flow data. Q/Q_0 = equilibrium flow/baseline flow.

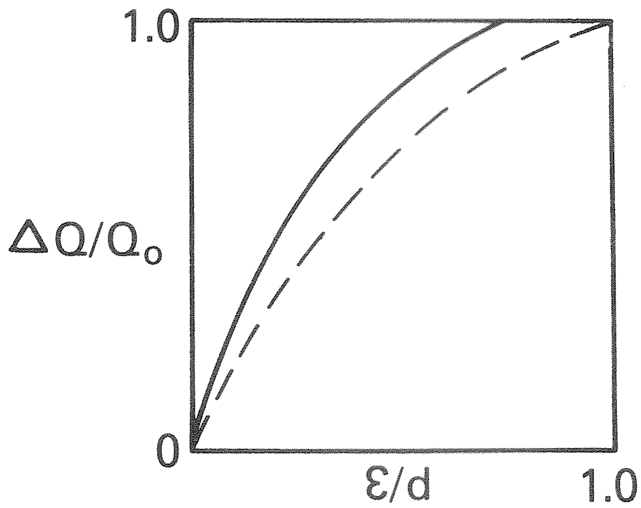

FIGURE 3
Dimensionless presentation of skin blood flow data for fixed value of d/D. ΔQ = blood flow decrease from baseline.

corresponding to each loading, the corresponding skin deflection, and the depth of the bone underlying the skin. This last parameter can be measured noninvasively with echo doppler equipment.

For a given bone depth ratio d/D, experimental data presented in this manner can be confined within a box bounded by the values 0 and 1 on both axes if we represent the reduction of blood flow as

$$\Delta Q/Q_0 = f(\epsilon/d) \qquad [4]$$

That is, blood flow cannot be reduced below zero ($\Delta Q/Q_0 = 1$), nor can we compress the tissue more than 100 percent ($\epsilon/d = 1$). Further, there is no reduction of flow for zero displacement (zero load), and there can be no flow when the tissue is fully compressed. Therefore, the experimental data would be expected to look somewhat like that shown in Figure 3.

THEORETICAL LOAD DISTRIBUTION UNDER INDENTOR

According to Timoshenko and Goodier (6, p. 371) the load distribution under a flat rigid cylindrical indentor resting on a flat, semi-infinite, homogeneous elastic body can be expressed mathematically as

$$q = (P/2\pi a^2)(1/\sqrt{1-(r/a)^2}) \qquad [5]$$

where q is the local loading (force per unit area), P is the total load applied to the cylinder, r is the radial distance from the centerline to the point in question, and a is the radius of the cylindrical indentor. It can be seen from Equation [5] that the pressure at the center of the indentor ($r=0$) is exactly one-half of the average pressure over the contact surface, whereas the loading at the outer rim ($r=a$) becomes infinitely large. A plot of this distribution is shown in Figure 4.

The theoretical model in Figure 4 is of course an approximation to the real situation in several respects. First, it is assumed that the elastic material (flesh) is homogeneous; second, the theory does not account for the presence of underlying bone or for curvature of the contact surface. However, the approximation is probably a good one for estimating loads at the center of the indentor, particularly since the inhomogeneity of the tissues is primarily in the axial direction normal to the layers of skin, subcutaneous tissue, etc. The general nature of the radial distribution curve seems to be somewhat realistic in that we noted a distinct reddening of the skin surface at

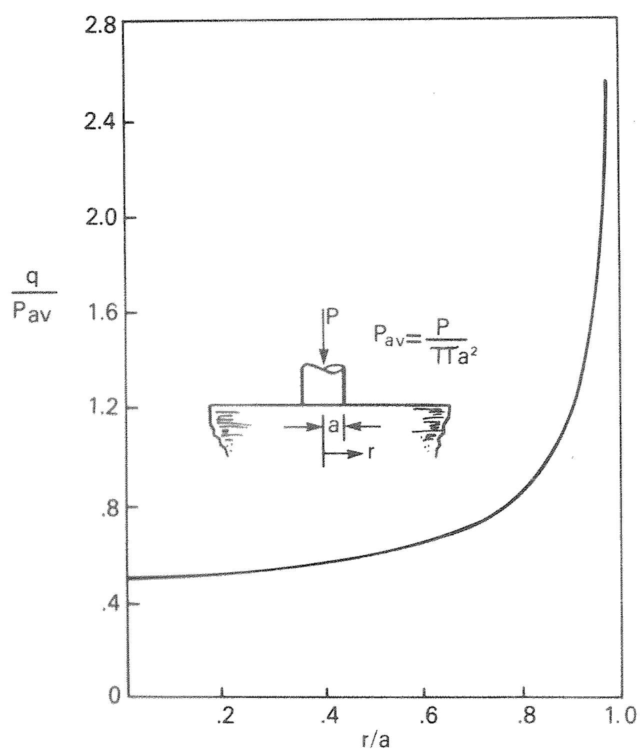

FIGURE 4
Theoretical load distribution under flat, rigid, cylindrical indentor. q = local loading; P = total load applied to cylinder; r = radial distance from centerline to point in question; a = radius of indentor.

the location of the outer edge of the indentor, despite a definite machined curvature of the sharp edge there.

With the same theoretical analysis, it is also possible to calculate the shape of an indentor that would produce a uniform load distribution. In fact, the shape can be expressed mathematically (6, p. 368) as

$$w = [4(1 - v^2)qa/\pi E] \times \int_0^{\pi/2} \sqrt{1 - (r/a)^2 sin^2 \psi} d\psi \quad [6]$$

where E is the elastic modulus, w is the local displacement from a flat surface at the radial position r, a is the radius of the indentor, q is the desired uniform loading, and ν is Poisson's ratio for the material. Note that the entire quantity $4(1-\nu^2)qa/\pi E$ is a constant under the stated assumptions. This indentor shape is plotted in dimensionless form in Figure 5. It can be seen that the absolute dimensions w depend on the magnitude q of the desired uniform loading and therefore on the applied load, as well as on the elastic modulus E. This indicates that there is no fixed shape that will provide a uniform load distribution to the tissue for all applied loads or for all tissues. That is, the shape depends on the load (being flatter for lower loadings) and on E (being flatter for stiffer tissues). Thus, it is only with a deformable indentor that one might hope to achieve a consistent, nearly uniform loading.

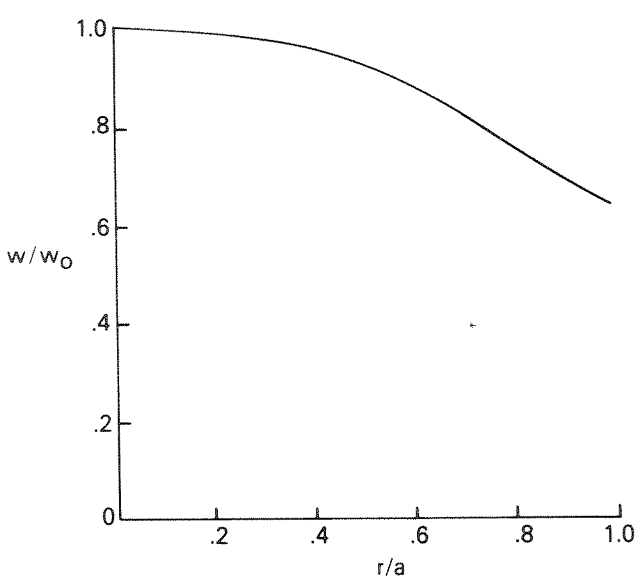

FIGURE 5
Theoretical indentor shape required to produce uniform load distribution. w = local displacement from flat surface at radial position r; a = radius of indentor; w_0 = value of w at r = 0.

TISSUE DEFORMATION

Regardless of the technique used to measure skin blood flow, it is relatively simple to measure skin displacement by using marks on the shaft that supports the load that will be applied with the indentor (Fig. 6).

We have used this simple technique to measure skin displacements beneath the indentor with loads applied vertically to the proximal femur of side-lying subjects (Fig. 7). We tested two able-bodied males (A.S. and H.O.) and two paraplegic males (J.S. and J.St.) and found that the "stiffness" of the tissues increased systematically with age, regardless of spinal cord injury. Thus, it can be seen from Figure 7 that *1)* the elastic behavior of the tissues is in all cases nonlinear, with the tissues getting stiffer (increasing slope) as the load increases and that *2)* the required load for a given displacement increases with age.

Such displacement information should always be obtained along with any skin blood flow measurements so that the data can be presented in the form $\Delta Q/Q_0 = f(\epsilon/d)$ as indicated by the dimensional analysis. This is probably a more meaningful representation than the usual dimensional plot of skin blood flow versus applied load. However, it should be noted that our displacement measurements were made immediately after each load was applied. This means that transient effects such as creep were not considered.

DISCUSSION

The reason for performing a dimensional analysis and presenting the results in the nondimensional form is that the results are then in a general form, thus permitting comparable data to fall on the same curve. This means that for a correctly selected group of subjects all data points should fall on the same curve (or surface). Thus, one might expect that data for all able-bodied subjects would fall on one curve, data for

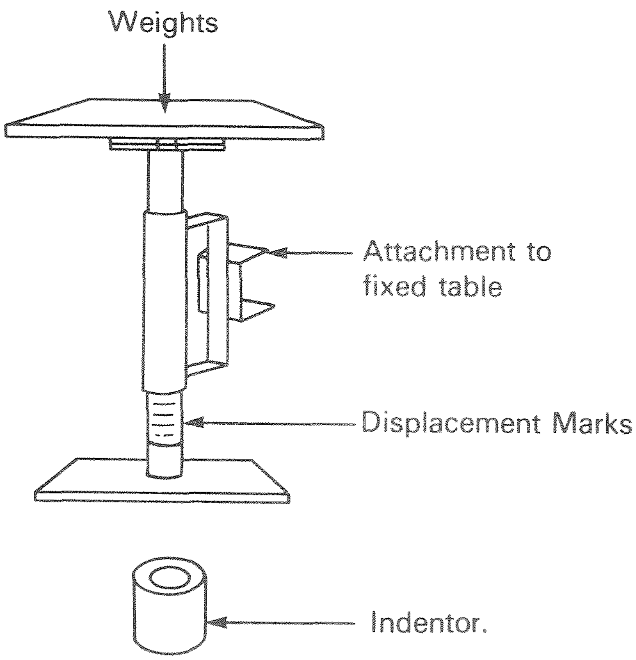

FIGURE 6
Schematic of loading fixture with displacement marks on shaft.

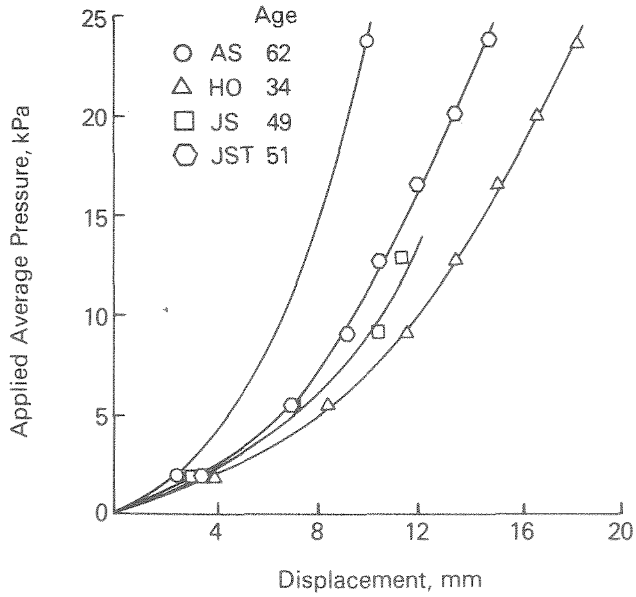

FIGURE 7
Skin displacements measured over proximal femur in 4 male subjects. Subjects without disabilities (*circle, triangle*); paraplegic subjects (*square, hexagon*).

all upper-neuron lesion spinal cord-injury patients would fall on another, and data for all lower-neuron lesion patients on another, unless these categories are inappropriate, or unless other factors are present which have not been accounted for in the analysis.

In any event, if the dimensional analysis has been correctly performed and if all of the primary factors have been included, then it should not be necessary to conduct large numbers of experiments to establish the desired curves. Furthermore, the scatter within the various groups should be small. Although others have apparently succeeded in separating some groups (e.g., geriatrics and paraplegics compared with young able-bodied subjects) on the basis of plots of blood flow versus pressure, the use of dimensional analysis and the associated nondimensional form of plots suggested here should improve separation and also permit the separation of more subtle groupings, such as those paraplegics who develop pressure sores compared with those who do not.

This procedure is a standard method of engineering and is the basis for all wind-tunnel testing of aircraft. That is, by dimensional analysis, one can use the results of model testing to predict the loads on the actual aircraft without the need to build and fly it. It seems reasonable that we should now avail ourselves of this technique in the systematic gathering of experimental data on alterations of skin blood flow produced by external loading, since the fluid and solid mechanics in both cases obey the same laws of dynamic similarity.

CONCLUSIONS

It seems from the dimensional analysis discussed here that skin blood flow measurements are best presented in the form of percentage of flow reduction as a function of the ratios of skin displacement to bone depth and bone depth to diameter of indentor. This would replace the customary presentation of blood flow as a function of applied pressure since the latter would be expected to depend, among other things, on the weight and obesity of the subject and on the elasticity of the tissues being compressed.

It has been shown that the load distribution under a flat, circular, rigid indentor is highly nonuniform, with extremely high local pressure at the rim of the indentor. An indentor shape can be designed to give a uniform load distribution, but this shape turns out to be a function of the applied load as well as the elastic modulus of the tissues. Thus, we would need a different indentor shape for each subject and for each load if we used a rigid indentor.

Measured skin displacements presented for four subjects indicate that the tissues are elastically nonlinear and that the stiffness of the tissues under compression increases with age, regardless of disability. Thus, a higher pressure on the skin indentor is required to produce the same degree of tissue compression in an older subject. It is this compression related to the bone depth (rather than pressure per se) that determines the compromise in skin blood flow produced by external loading.

Such displacement measurements can be obtained with surprising simplicity by using the movement of the load applicator shaft. We have shown that such measurements are important for the proper presentation and interpretation of skin blood flow measurements associated with the application of external pressure.

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