TRANSFERRING LOAD TO FLESH
PART VII. GEL LINER EFFECTS [a]

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ABSTRACT

A simulated gel-lined socket generally produces compressive and shear stresses in flesh proximal to the socket, when compared with an unlined socket. The gel liner acts to produce a more nearly uniform distribution in flesh, thereby reducing peak stresses near the load. However, stresses at locales remote from the loading point are increased. As these latter stresses are usually low in any event, an increase would seem acceptable. From the stress point of view a gel socket liner appears to be worthwhile.

The work presented consists of experimental measurements of compressure and shear stress within a simulated flesh model. Tests include variations in brim radius and in gel thickness. An unsuccessful attempt to determine combined stress is outlined.

INTRODUCTION

We are concerned with reducing or preventing incidents of trauma to soft tissue experienced by amputees and brace wearers. Specifically, we are interested in the process of load transfer from a prosthesis or brace to a subject, with the goal of minimizing the flesh stress so experienced to a value below damage thresholds.

This paper deals with the effects on flesh stress of lining a prosthetic socket with a material known as Spence-gel.[b] This gel is a formulation whose characteristics are similar to those of human flesh in terms of stiffness and viscoelastic properties. At issue is a practical question—is the gel a worthwhile improvement as compared to a conventional hard socket?

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[a] Based on work performed under VA Contract V101 (134)P–18.

[b] Registered trade name. A silicone formulation produced by the Dow Corning Corp. for the Stryker Corp., Kalamazoo, Mich.

The gel liner concept is well-known and has been advocated as a useful measure by workers at the University of Michigan (1). Socket pressure studies when a gel liner is employed have indicated a decrease in stump pressure peak values, which implies a corresponding decrease in stump compressive stress. However, pressure testing at discrete points at the socket-flesh interface, while most useful, tells little of compressive stress distribution within the flesh and nothing whatever of the existent shear and tensile stresses in the stump. Therefore, while the gel-liner concept investigated is well known, it is believed that the results given here are fresh.

While the techniques employed here are briefly outlined, little background is supplied in this paper. Those seeking a more complete discussion of procedure and outlook may examine Reference 2, a qualitative discussion of stress trends in non-rigid materials; Reference 3, an analytic treatment of compressive stress in flesh; Reference 4, an analytic treatment of shear stress in flesh; Reference 5, an analytic treatment of contact curvature effects upon flesh stress; Reference 6, an experimental technique capable of modeling flesh stresses; and Reference 7, an experimental treatment of brim radius effects.

METHOD OF PROCEDURE

A flesh substitute in a thin slab form is marked with coordinate lines and then inserted between rigid walls. A load (*constant at 500 gm. or 1.1 lb.*) is applied to a member bearing on the model flesh and within its plane. Deflection of the coordinate lines is photographed and becomes raw data. Calibration of the model flesh permits conversion of the deflection information to corresponding stress results. Thus, the squeezing together of the coordinate lines is interpreted as compressive stress, and the departure from a right angle at the intersection of coordinate lines is interpreted as shear. Combined stress calculations followed the classic construction (Mohr's circle) detailed in any elementary text on stress analysis (8).

The particular section examined corresponds to the dashed rectangle in Figure 1—the anterior brim of an above-knee socket. A series of socket brims of varied radii (1 in., ½ in., ¼ in., and ⅛ in.) attached to mock socket sections are covered with varied gel thicknesses (¼ in., ⅛ in., and 0). The test load, applied as shown, corresponds to a forward thrust of the hip. That is, while the test load is applied as a force external to the socket brim segment, it is precisely equal, in its effect, to an inertial load experienced in accelerating the lower limb in the swing phase. The femur is remote from the loading point; i.e., the stump is considered fleshy.

Data-handling techniques have been changed in one significant

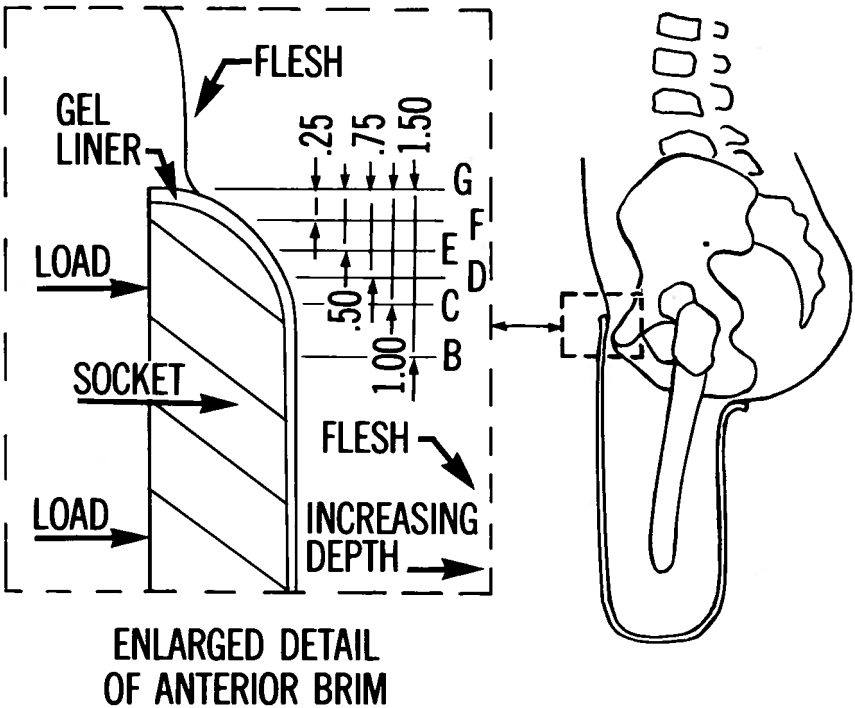

FIGURE 1.—Schematic drawing of the socket section simulated in the test. Area within dashed rectangle is tested. Enlarged section shows planes along which stresses are computed. Every following figure refers to a given plane defined in the rectangle.

aspect with respect to prior work. As we were concerned with the large scatter apparent in previous experimental results, a number of replica runs have been conducted in these gel effect studies. Each test condition has been conducted two or three times and the resulting values averaged. The effect is to smooth the results. Therefore, the reader should note that each plotted point in the accompanying graphs is not a true data point, but rather the average of a number of scattered points.

RESULTS

The results, in the form of compressive and shear stress within the model flesh versus depth of flesh (from the socket wall inwards) are given in Figures 2–25. Each figure represents the stress along a designated (see legend) horizontal plane, located at varying distance from the brim. (Plane B, 1.5 in.; C, 1 in.; D, .75 in.; E, .50 in.; F, .25 in.; G, at brim level). Brim radius, a major test parameter, is represented by different groups of figures. Thus Figures 2–7 correspond to the

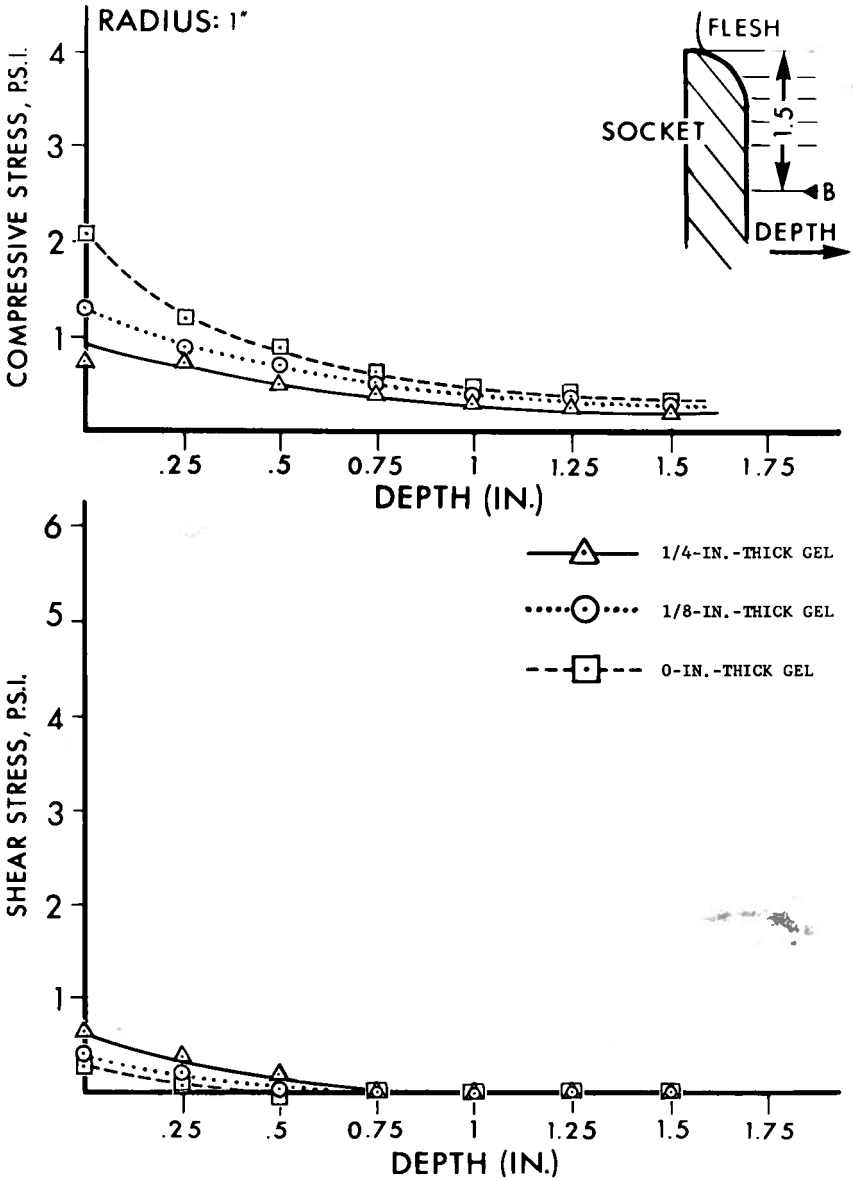

FIGURE 2.—Test results, 1 in. brim radius, plane B.

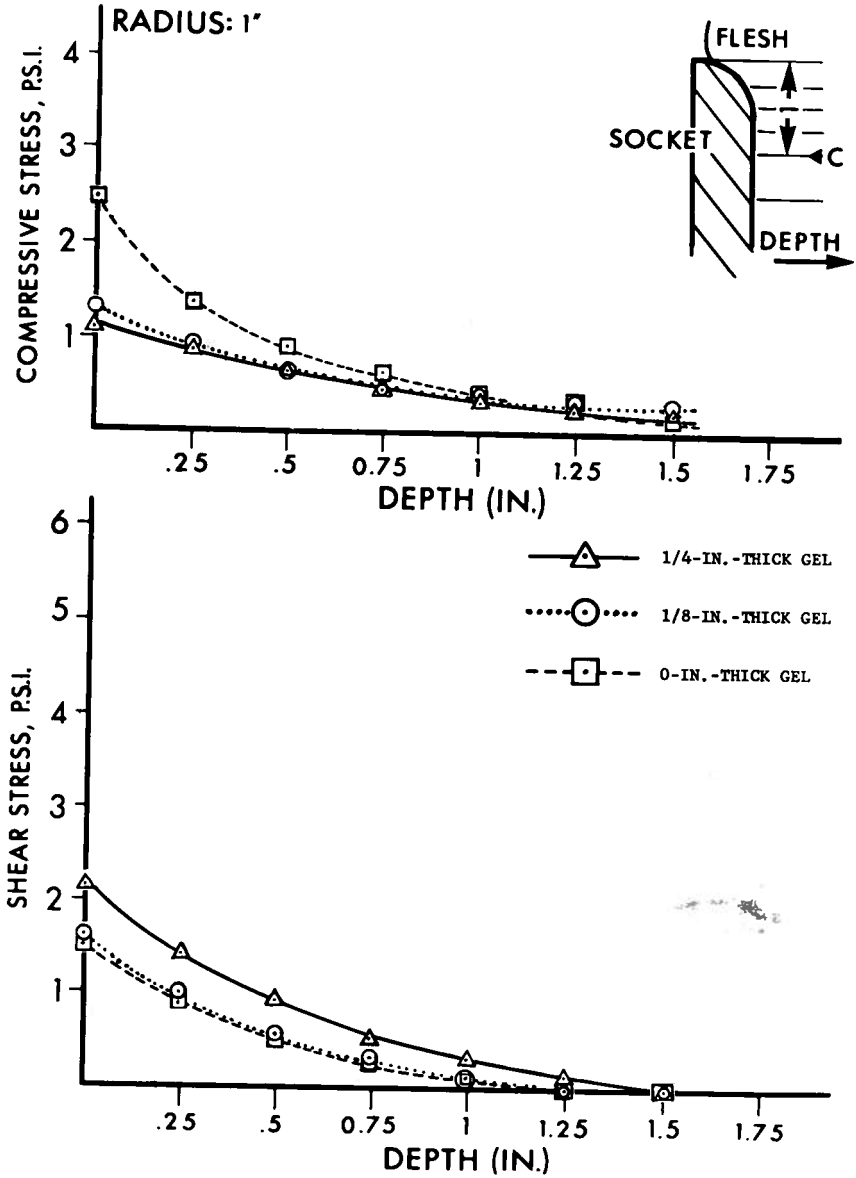

FIGURE 3.—Test results, 1 in. brim radius, Plane C.

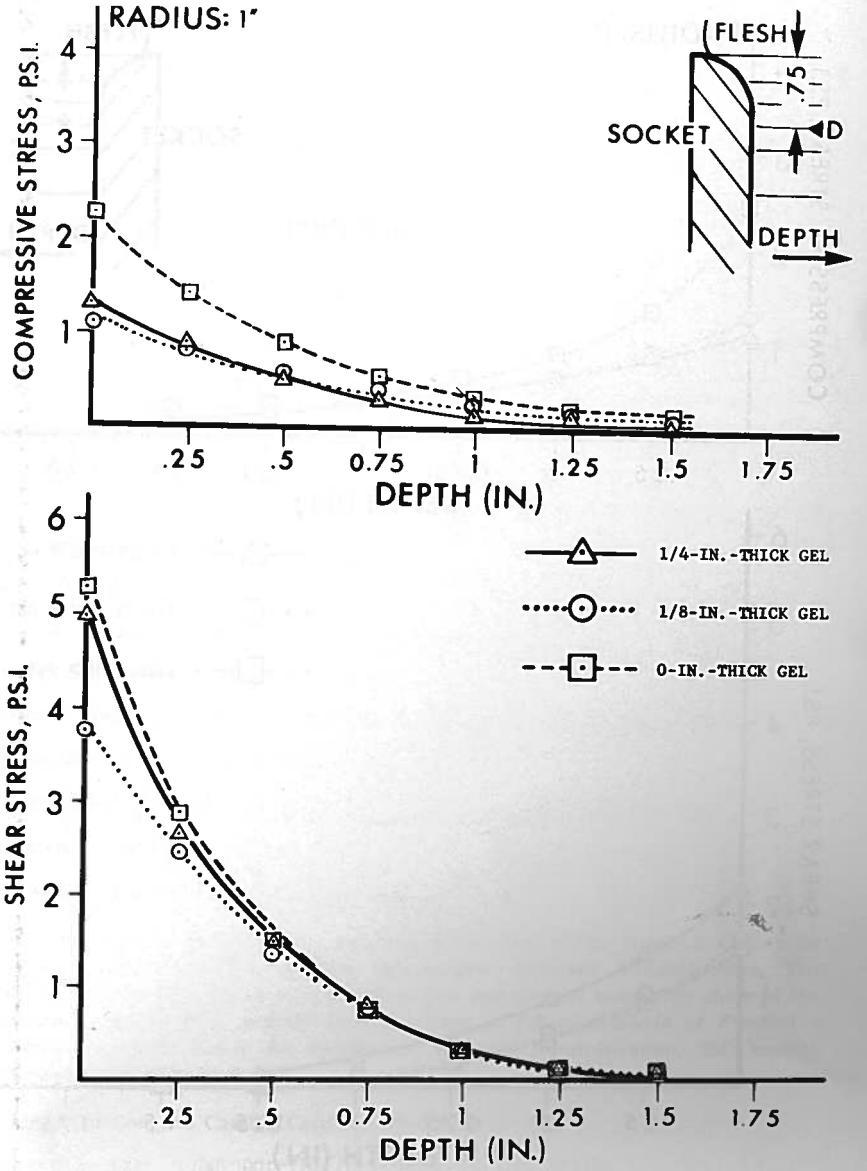

FIGURE 4.—Test results, 1 in. brim radius, Plane D.

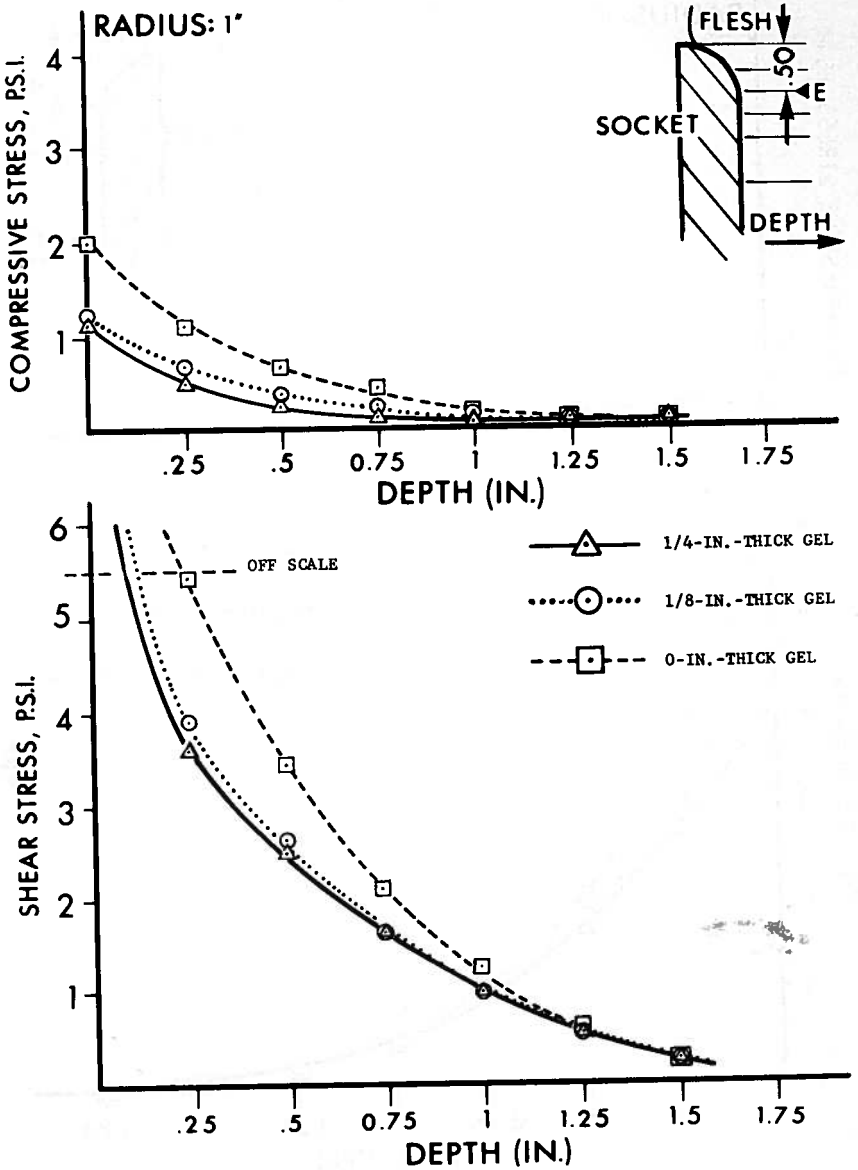

FIGURE 5.—Test results, 1 in. brim radius, plane E.

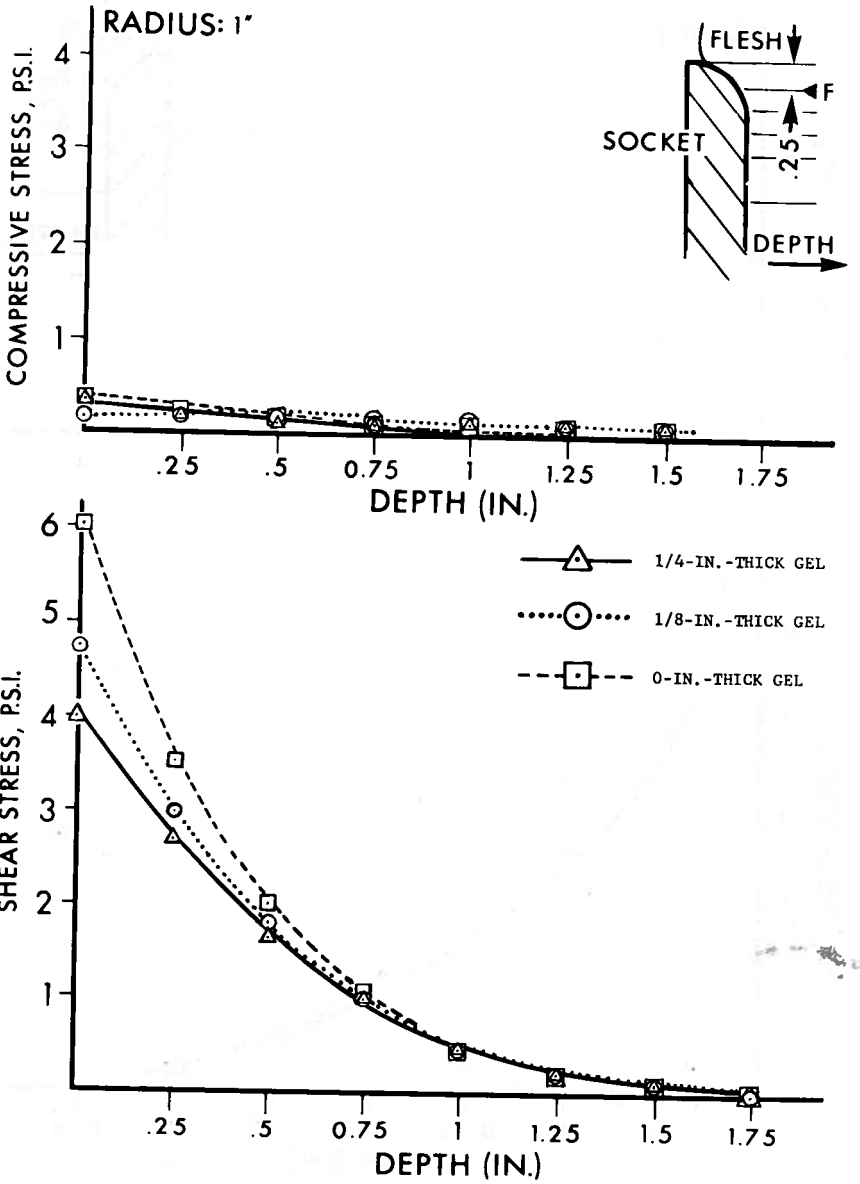

FIGURE 6.—Test results, 1 in. brim radius, plane F.

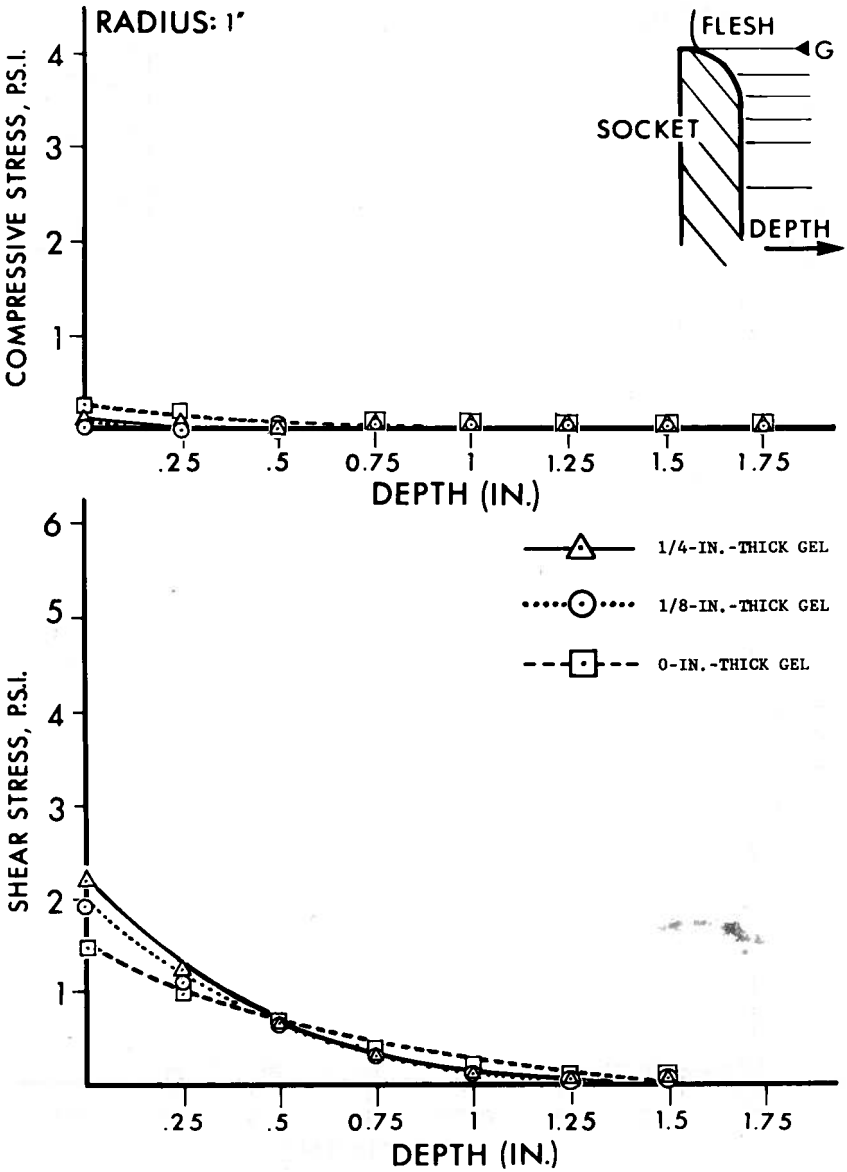

FIGURE 7.—Test results, 1 in. brim radius, plane G.

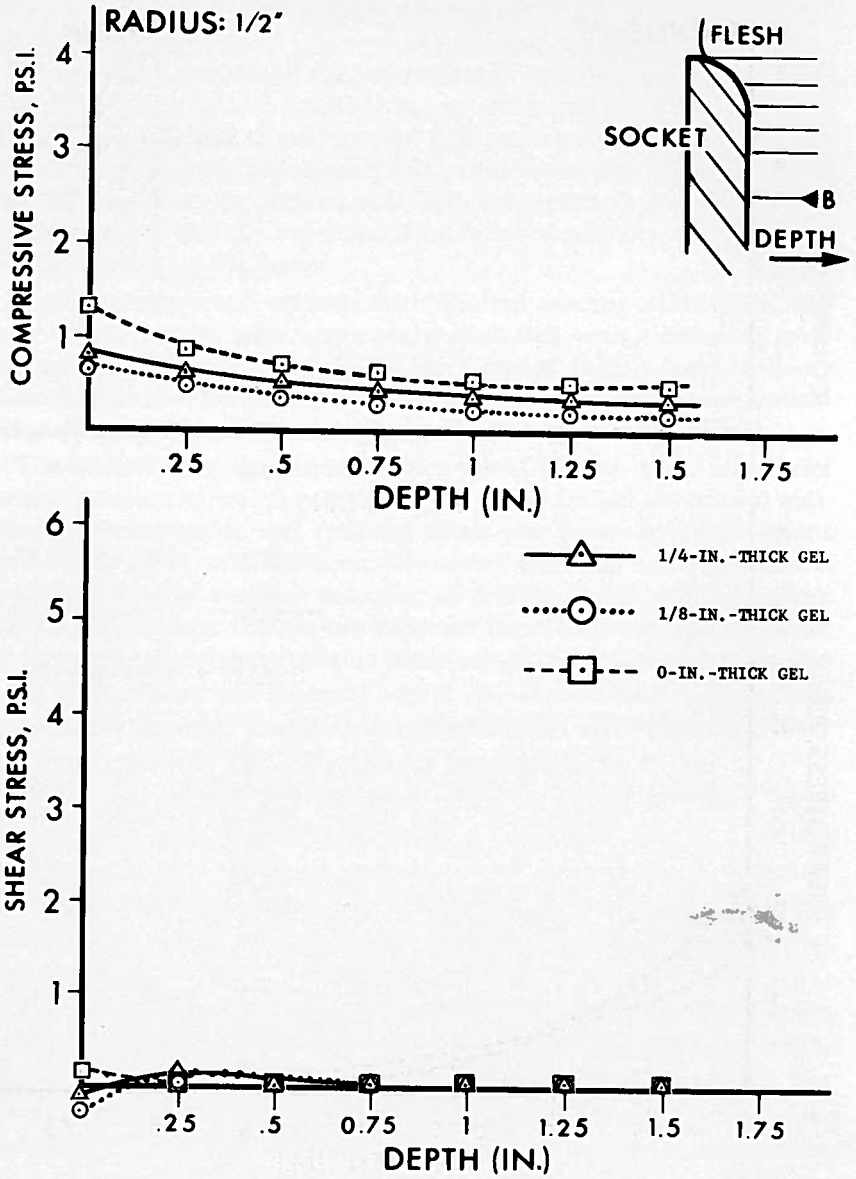

FIGURE 8.—Test results, ½ in. brim radius, plane B.

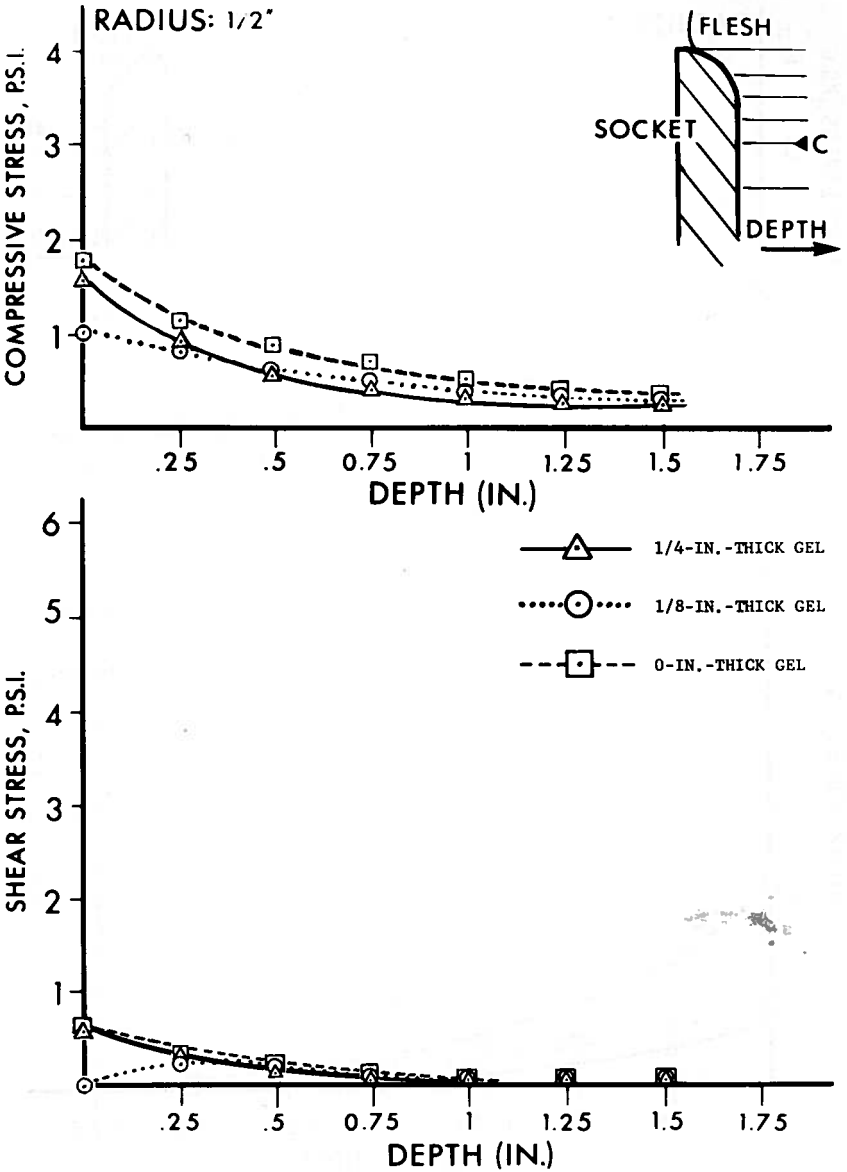

FIGURE 9.—Test results, ½ in. brim radius, plane C.

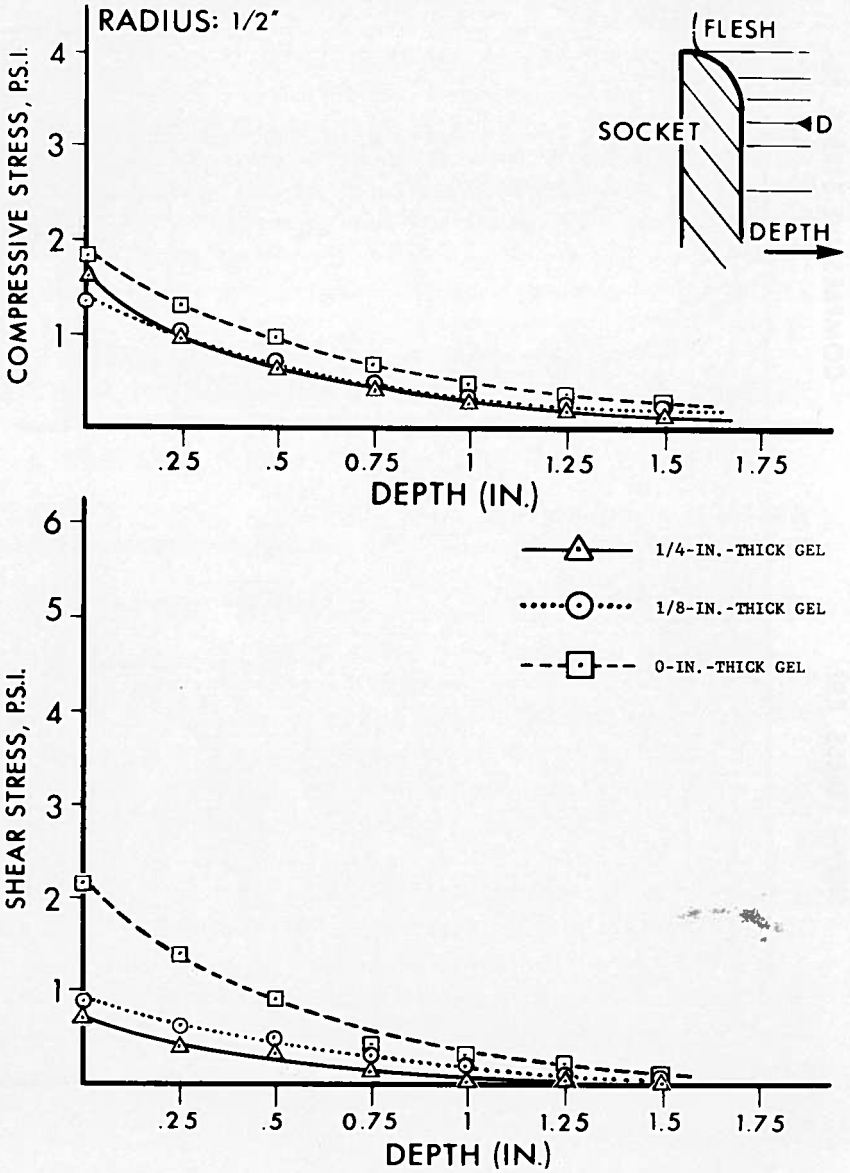

FIGURE 10.—Test results, ½ in. brim radius, plane D.

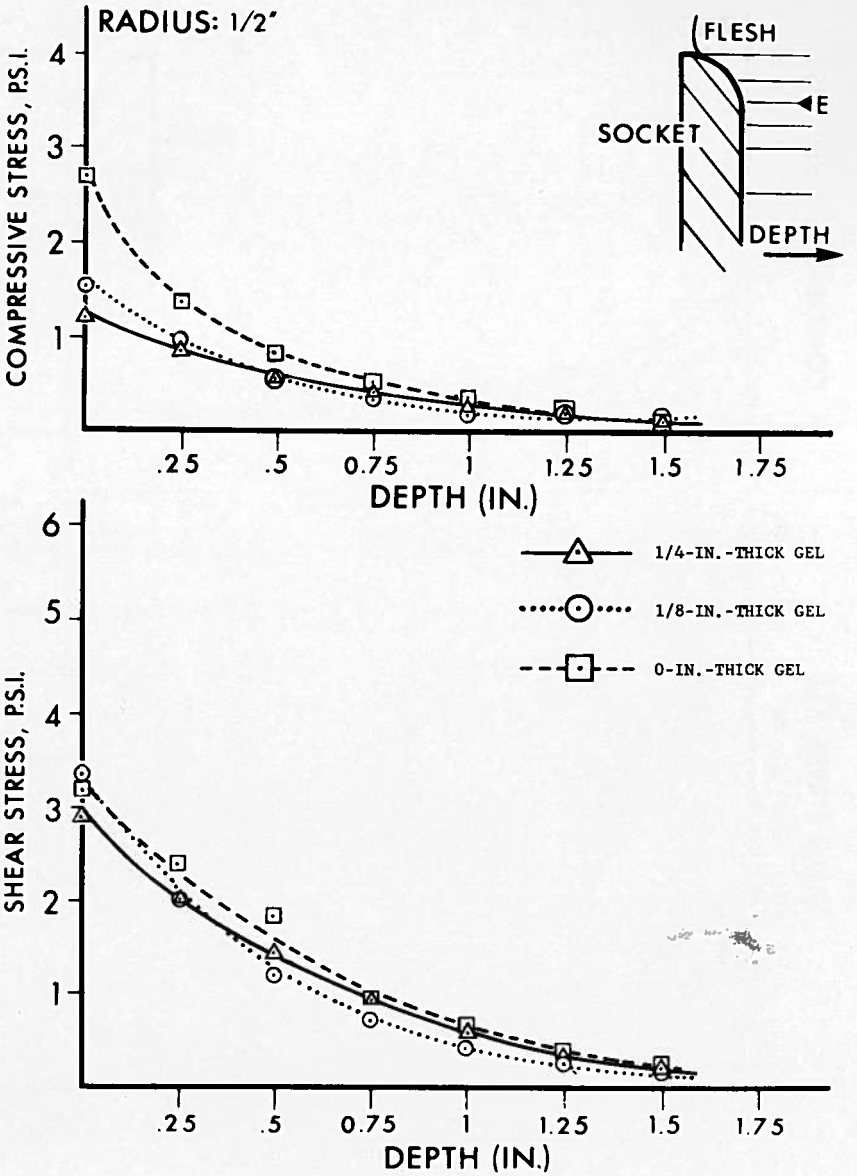

FIGURE 11.—Test results, ½ in. brim radius, plane E.

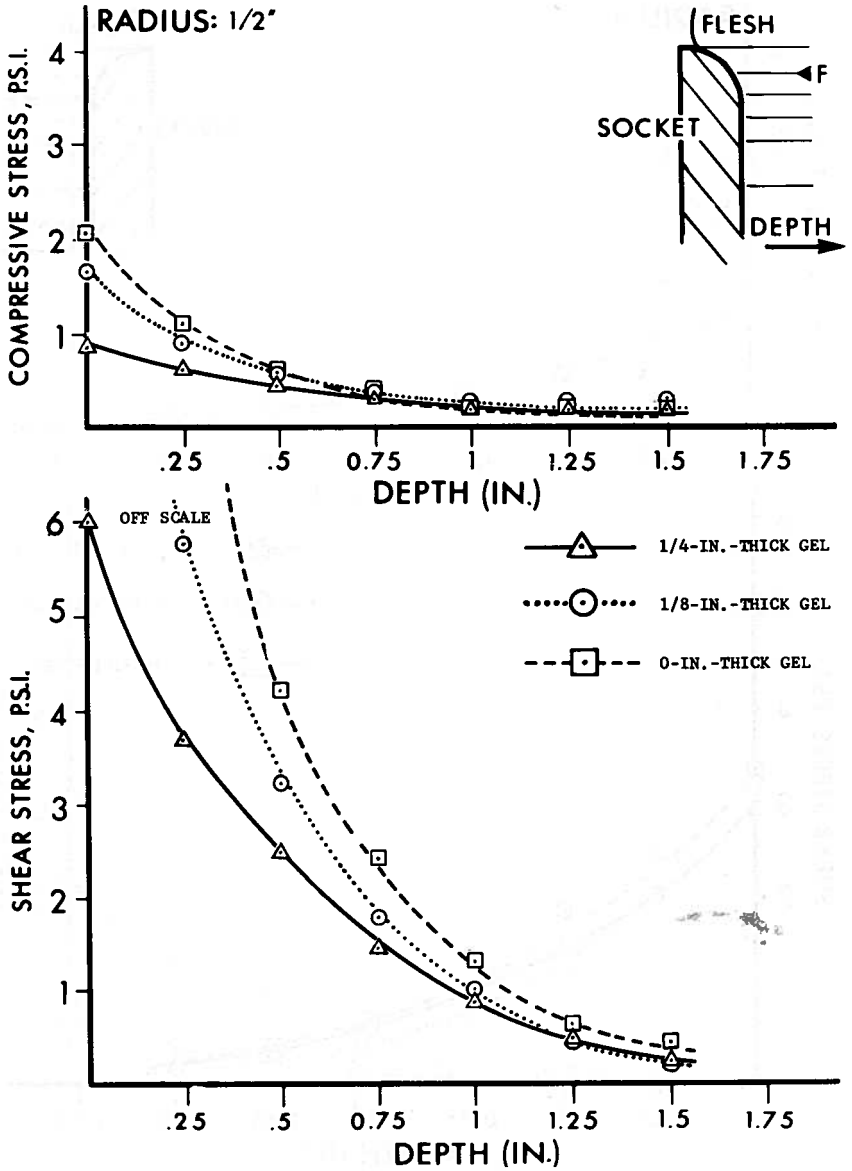

FIGURE 12.—Test results, ½ in. brim radius, plane F.

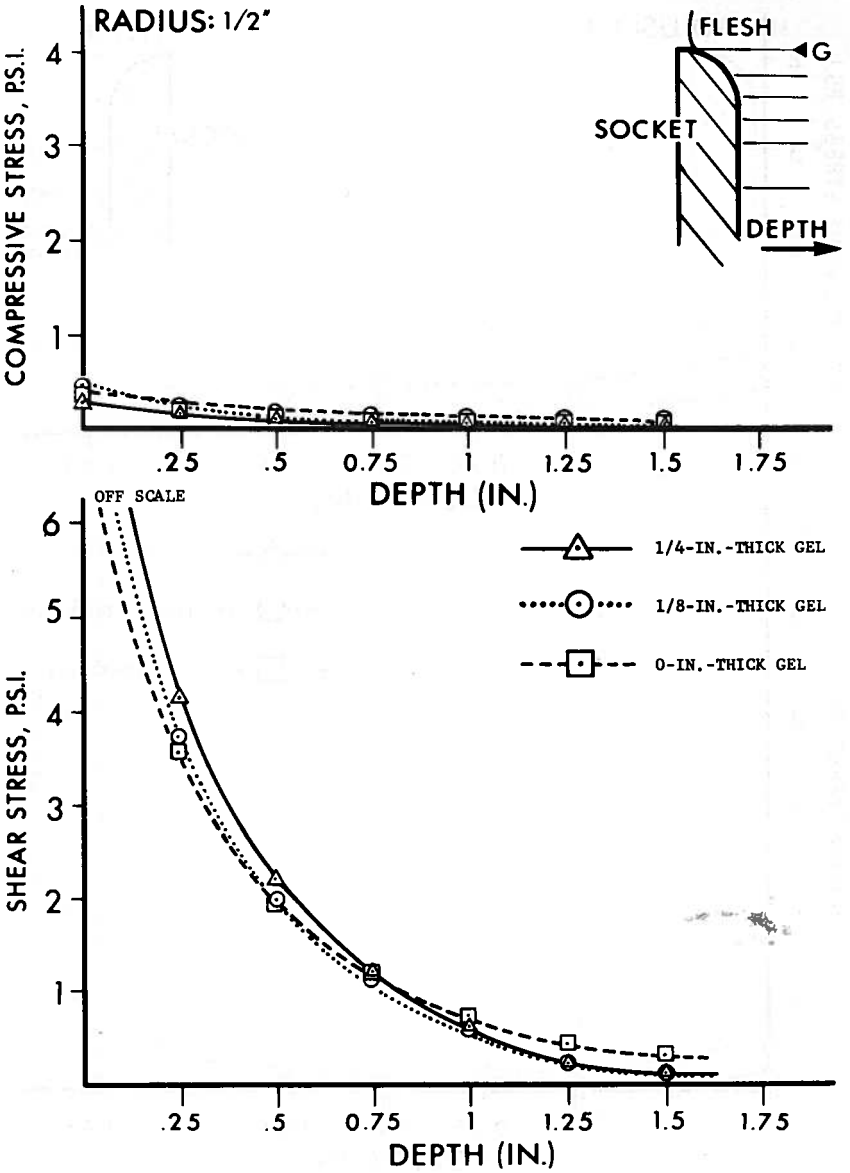

FIGURE 13.—Test results, ½ in. brim radius, plane G.

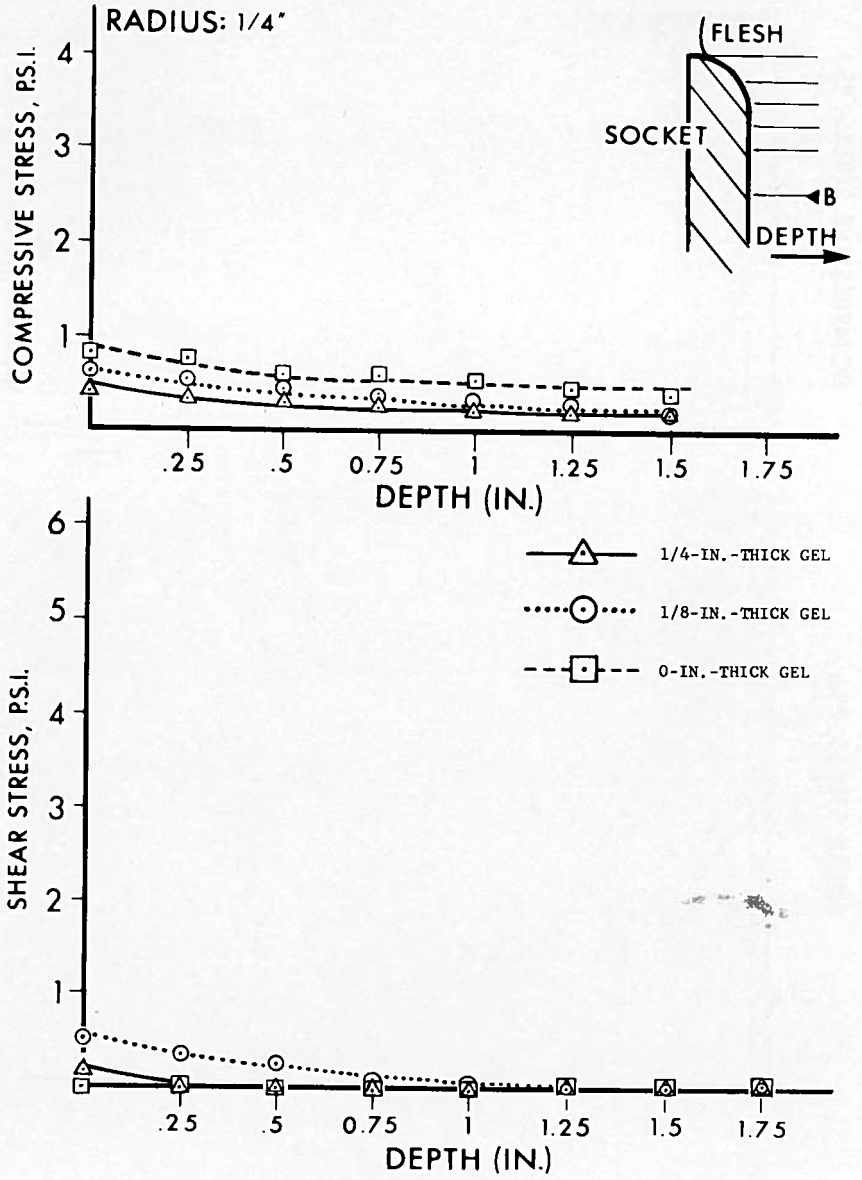

FIGURE 14.—Test results, ¼ in. brim radius, plane B.

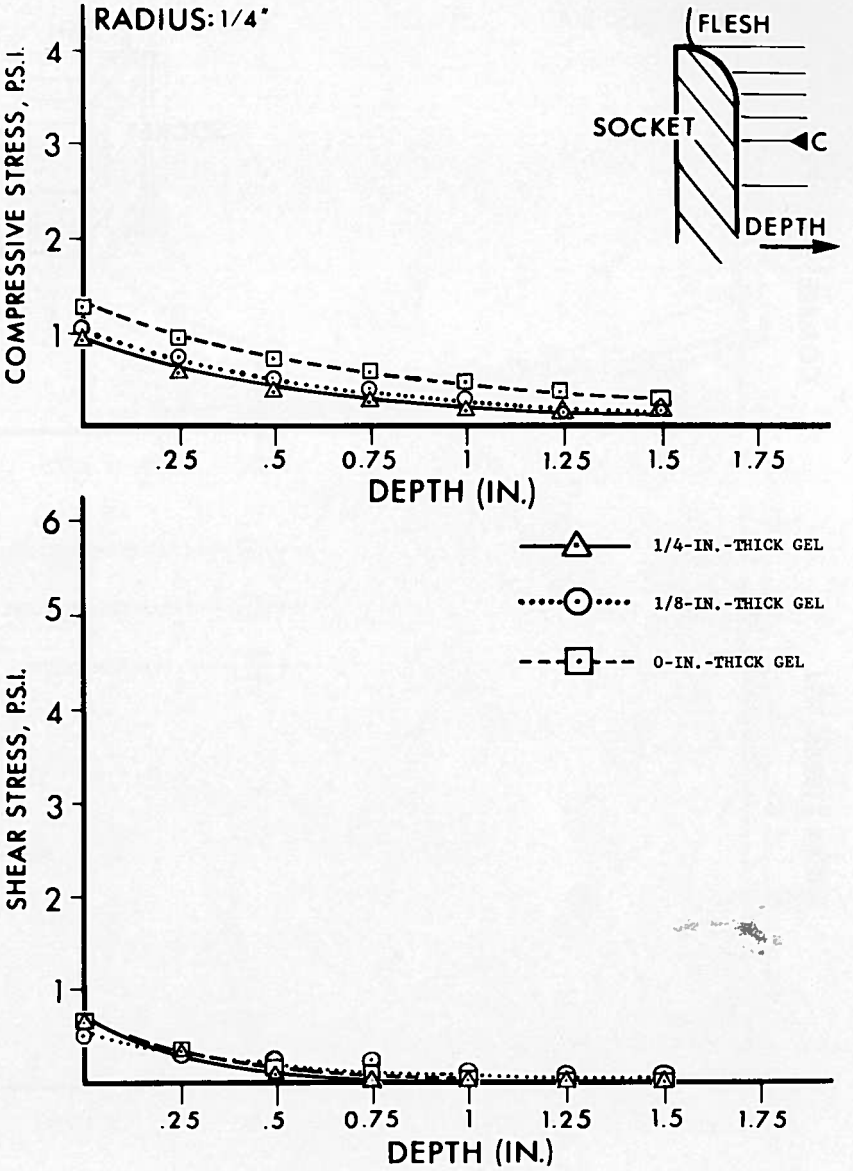

FIGURE 15.—Test results, ¼ in. brim radius, plane C.

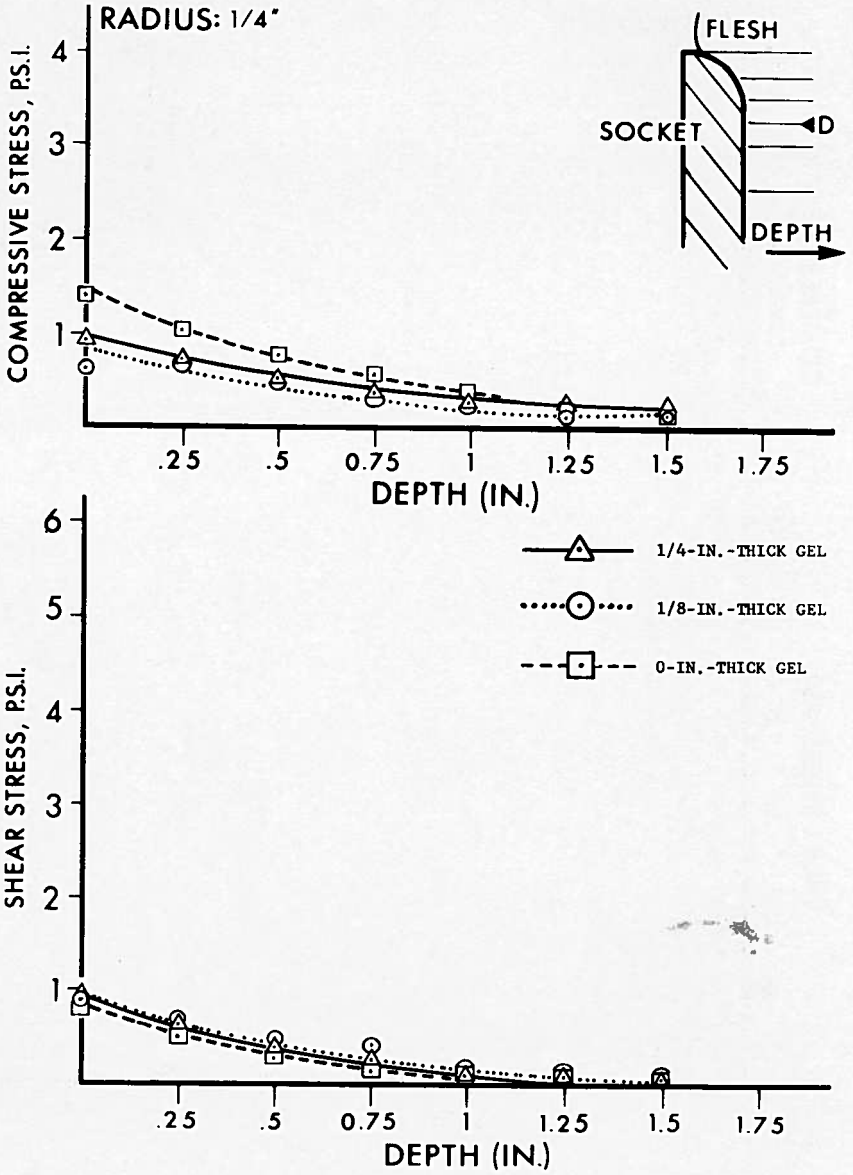

FIGURE 16.—Test results, ¼ in. brim radius, plane D.

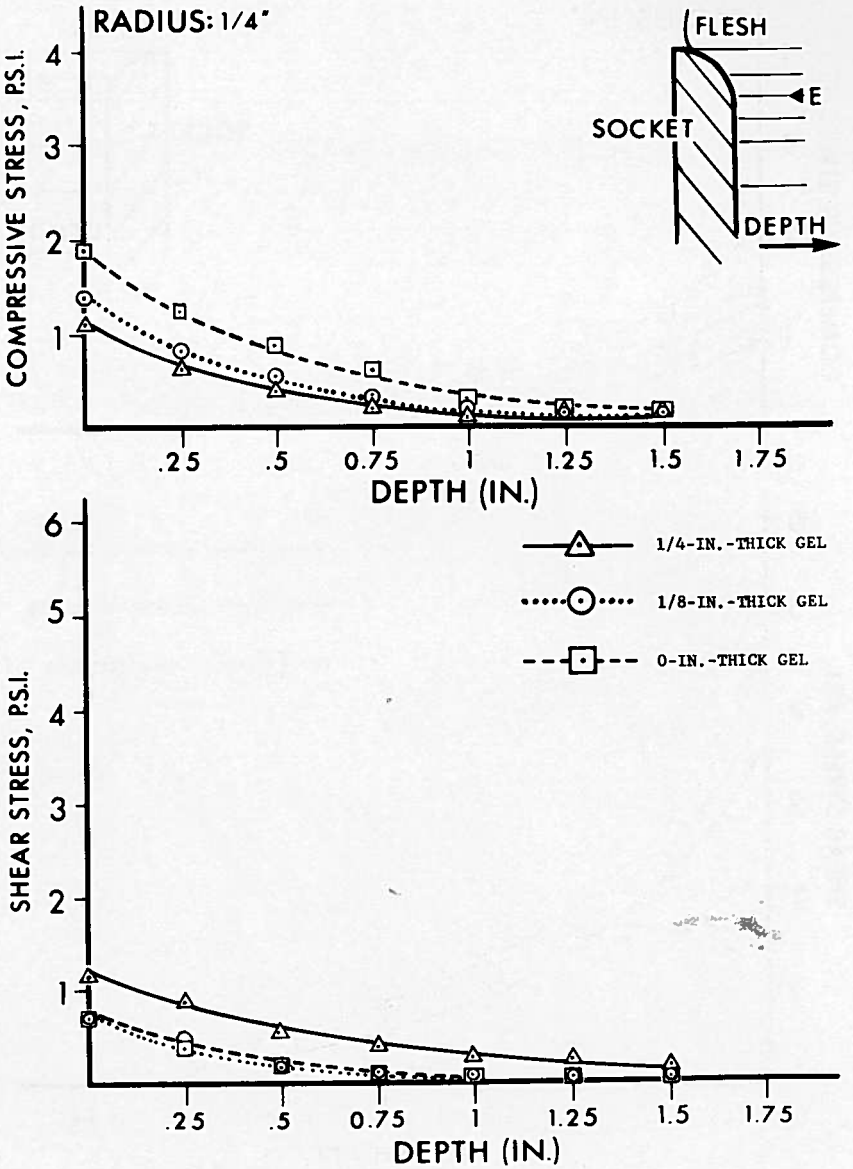

FIGURE 17.—Test results, ¼ in. brim radius, plane E.

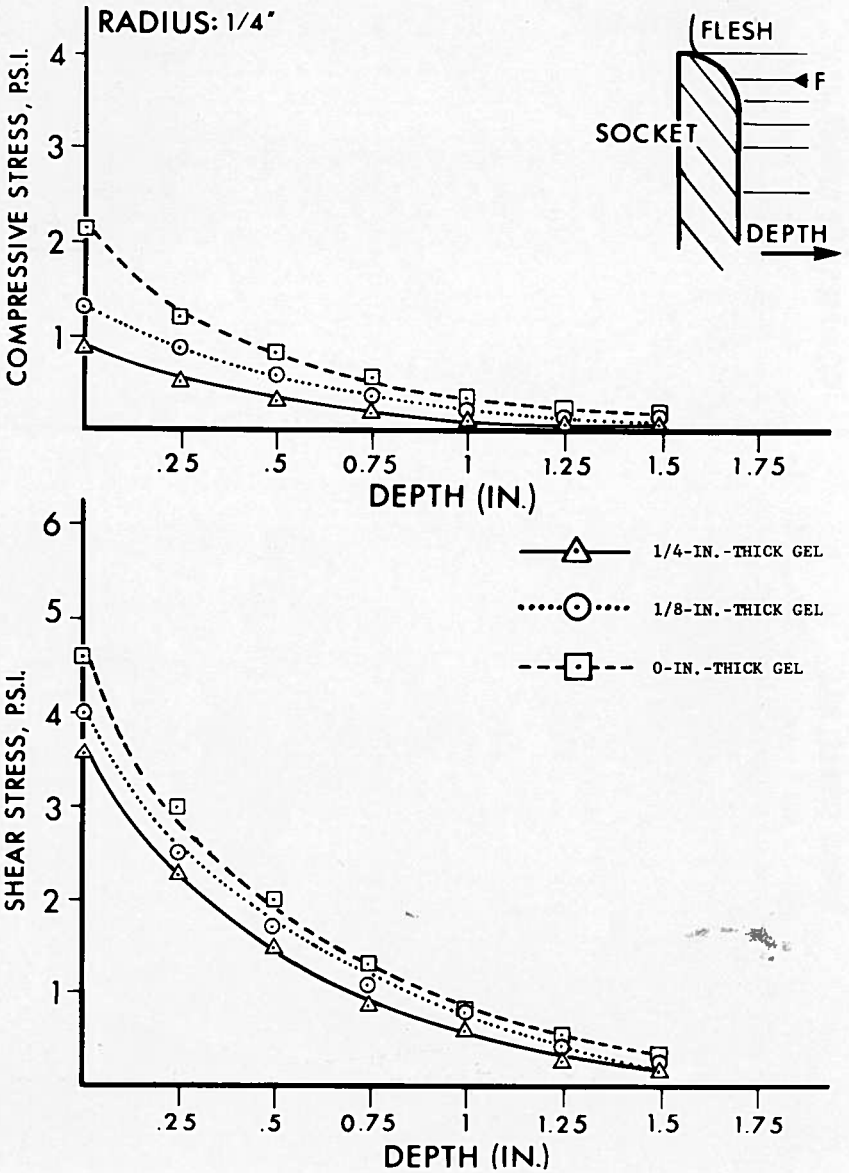

FIGURE 18.—Test results, ¼ in. brim radius, plane F.

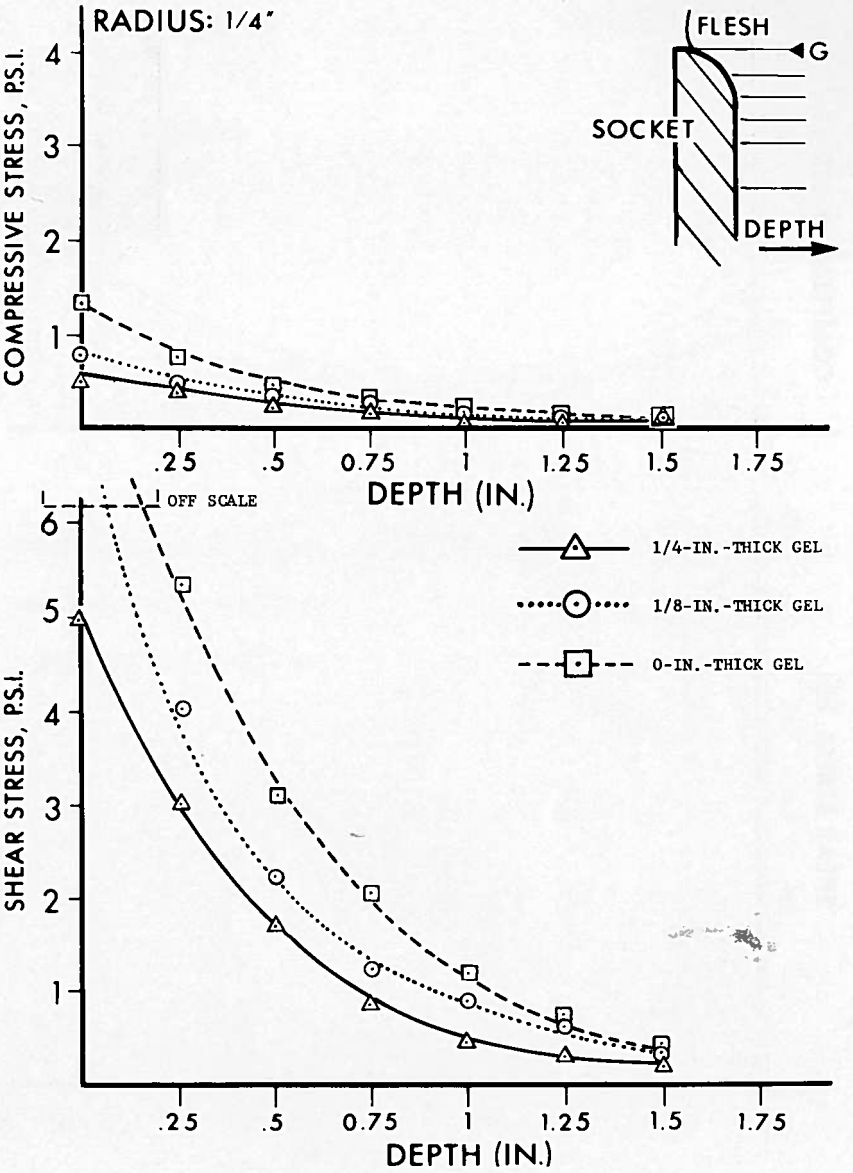

FIGURE 19.—Test results, ¼ in. brim radius, plane G.

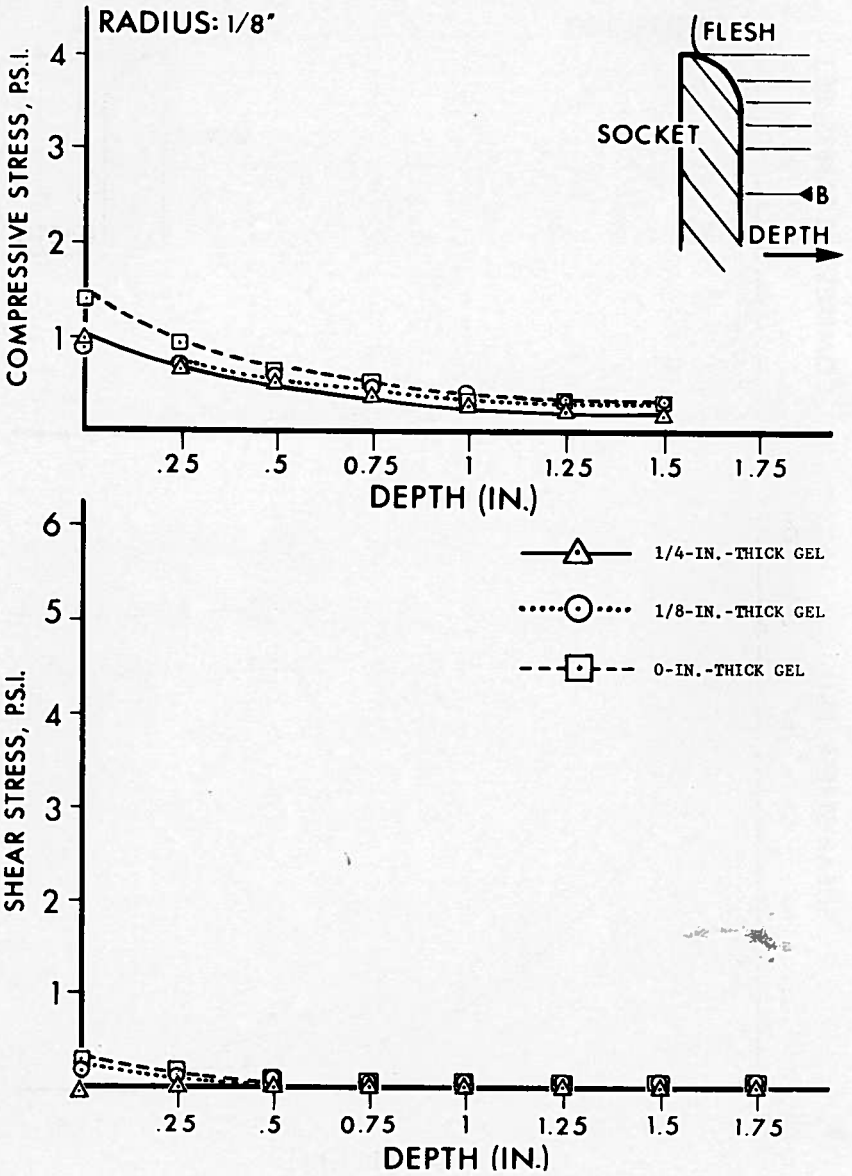

FIGURE 20.—Test results, 1/8 in. brim radius, plane B.

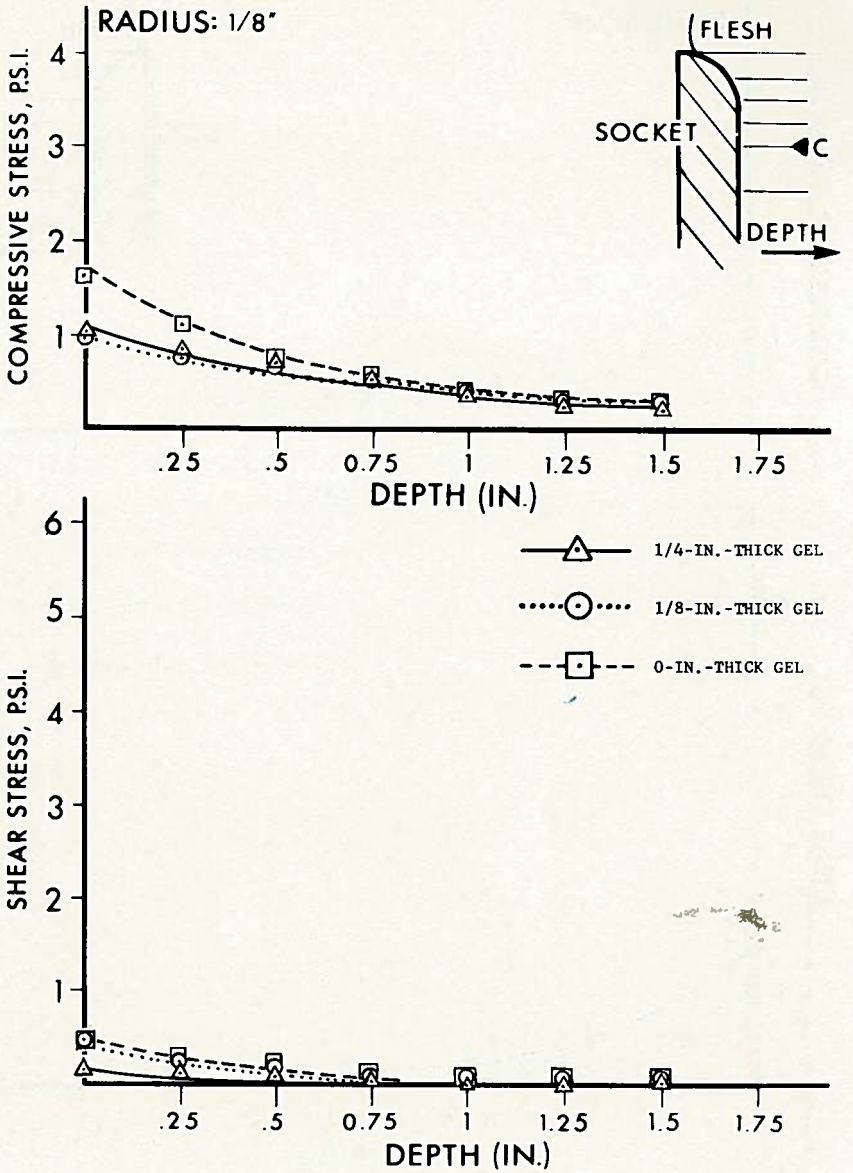

FIGURE 21.—Test results, 1/8 in. brim radius, plane C.

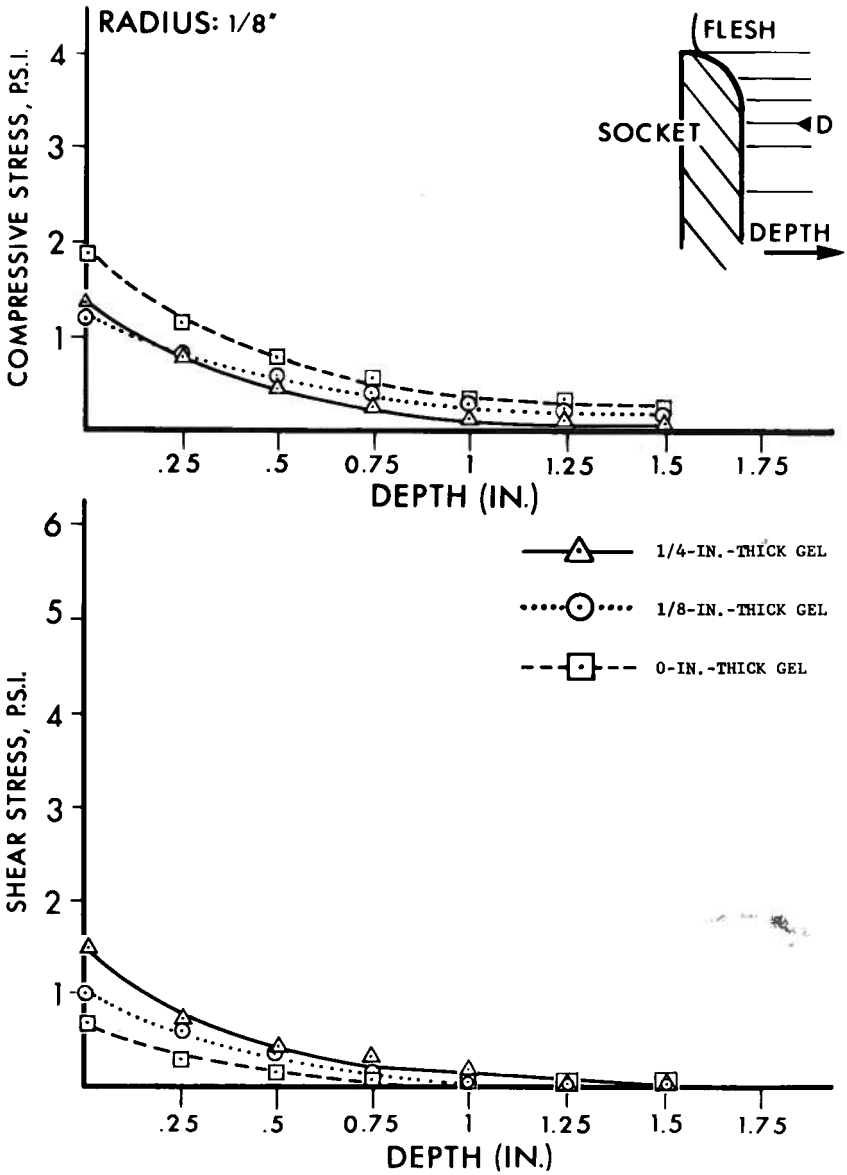

FIGURE 22.—Test results, 1/8 in. brim radius, plane D.

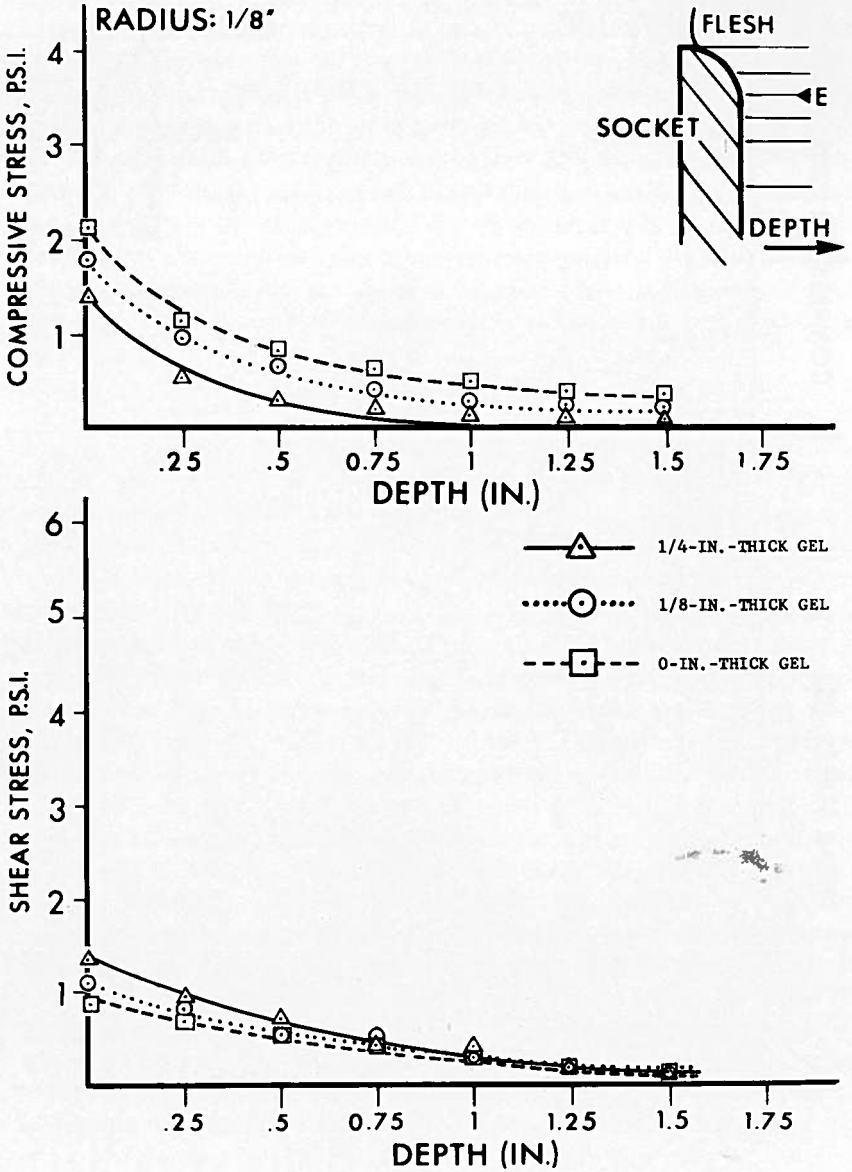

FIGURE 23.—Test results, 1/8 in. brim radius, plane E.

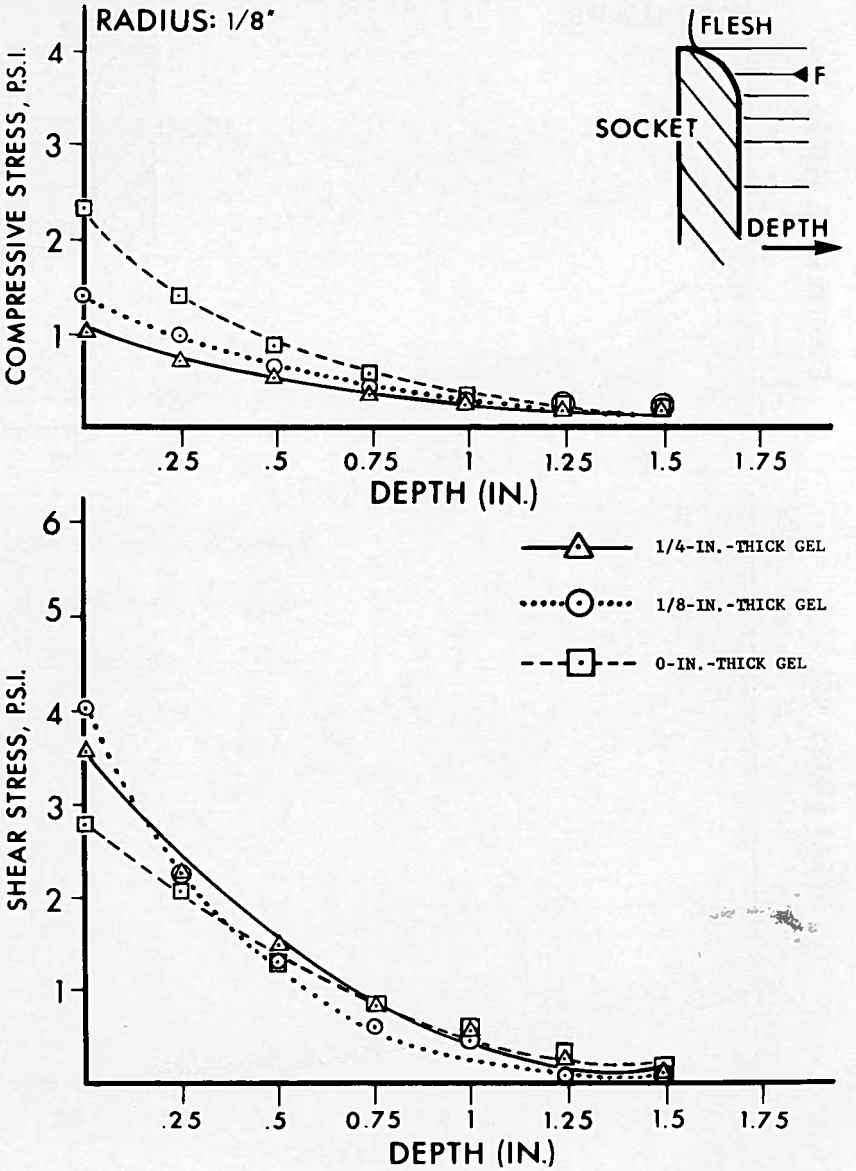

FIGURE 24.—Test results, 1/8 in. brim radius, plane F.

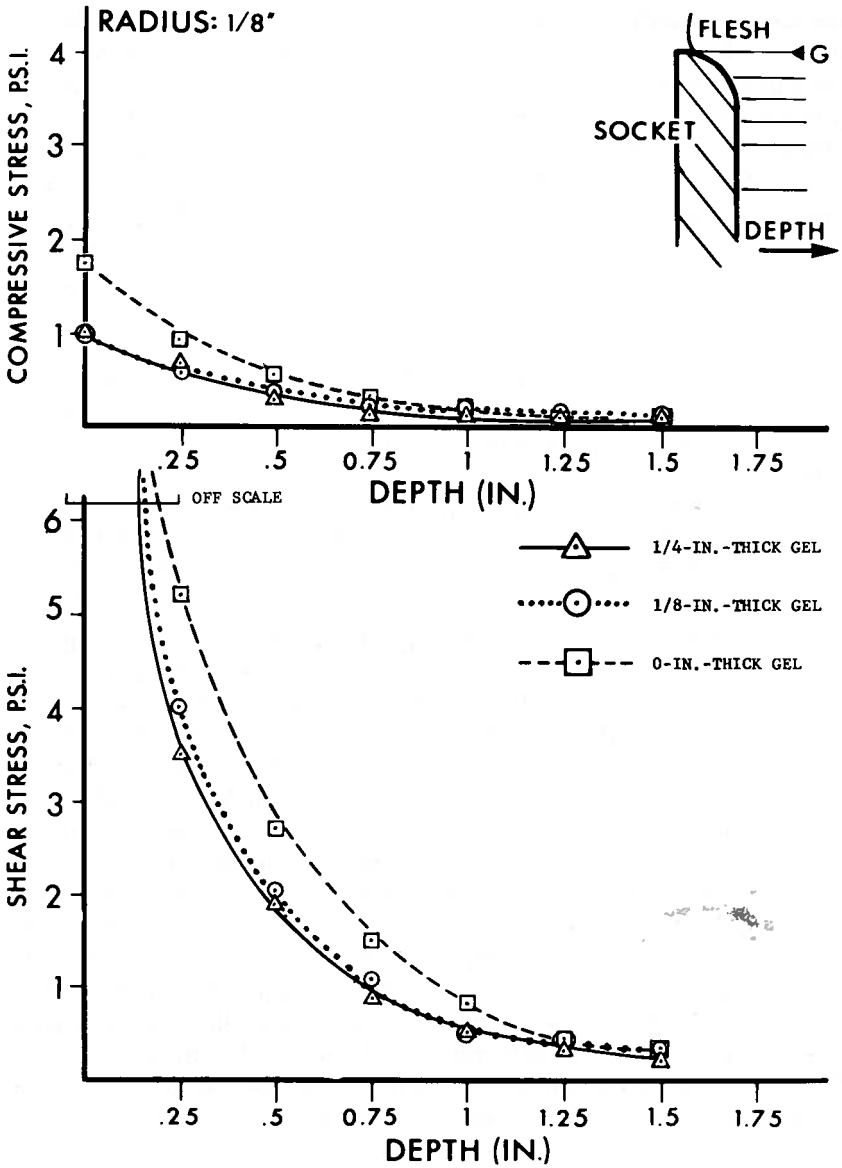

FIGURE 25.—Test results, 1/8 in. brim radius, plane G.

largest test radius (1 in.), Figures 8–13 to half the largest radius (½ in.) and so on through a radius of ⅛ in. In each case brim radius is defined as that value which existed before the addition of the gel cover. Two thicknesses of gel are tested; each plot gives the appropriate symbols and the resulting stresses. Occasional wild points are presented without comment, for example Figure 9, 0 in. depth, ⅛ in. thick gel.

Consider the large radius results (1 in.), Figures 2–7. Note that stress is always a maximum at the skin surface (0 in. depth) and decreases steadily moving towards the femur. Shear stress increases as the level examined is taken closer towards the brim reaching a maximum within the flesh in contact with the brim. Compressive stress peak values are experienced deeper into the socket and decrease within the brim locale. The above characteristics are not influenced by the presence or the amount of gel cover. The gel acts to reduce significantly the stress—both compressive and shear—when the peak stress level is high, say at about the 4 p.s.i. level in the case of shear and at the 1 p.s.i. level in the case of compressive stress. When the peak stress level is moderate—at, say half the above values, the gel cover either does little, or actually serves to increase stress values. See Figures 2 and 3 shear values.

As the brim radius is made small, stress factors in the brim neighborhood are altered considerably. This change may be seen most readily by comparing the results obtained from extreme radii values of 1 in. versus ⅛ in. (Fig. 20–25). Deep into the socket on planes B and C, there is little difference between results; i.e., Figure 2 is similar to Figure 20 and Figure 3 is similar to Figure 21. Moving upwards to the D plane the large radius socket brim is sufficiently close to plane D (because of its size) to create considerable local shear stress (Fig. 4) as compared to the small radius values (Fig. 22). Plane E results, Figures 5 and 23, show the same trend magnified. The large radius shear value has "blown up" as the plane examined is taken at the brim, while the small radius values, at the same plane, remain moderate. Succeeding figures, especially Figures 7 and 25, illustrate this decay of the large radius shear stress, once well into the "turn" of the brim corner and the corresponding growth of the small radius shear stress as the plane of observation (G) is now taken within its own, smaller sphere of influence. Note that the small radius shear stress peaks after the turn—in contrast to the large radius peak within the turn itself. Given a small radius, the gel coat is not particularly effective in reducing shear stress, except in the case of peak shear stress (Fig. 25). However, the gel coat is useful in reducing compressive stresses, achieving roughly the same percent of reduction with a small or large radius brim.

Examination of intermediate radii results (Fig. 14–19 and 8–12) reveals no new trend. Again the gel coat is seen effective in reducing

shear stress only when the peak shear stress is large (Fig. 12 and 19). Given a moderate level of shear stress (Fig. 17) the gel coat may actually increase the local shear stress. Where compressive stresses are concerned, the gel coat always acts to reduce stress value without the "threshold" effect noted in shear.

Attempts to determine combined stress values (stresses arising from the combination of normal and shear stress are called combined stress), from those values of compression and shear given above, were not successful. Conventional Mohr circle construction led to nonsense results which are not given. It is believed that this difficulty comes from an implicitly assumed homogeneous material with isotropic characteristics necessary to utilize the Mohr method, whereas both the real and model flesh are clearly non-isotropic. See Discussion.

DISCUSSION

The effects of the gel coating are to reduce compressive stresses in almost all circumstances (without respect to brim radius) and to reduce shear stresses, where the initial shear stress level is very high. Should the initial (uncovered) shear stress level be low, the gel coat may well act to increase the local shear stress level.

The gel coat may be viewed as a leveling agent, reducing stress peaks at the price of higher shear stress at remote locations. If we view soft tissue trauma as linked to high stress, then a gel socket coat is desirable. While it is true that the gel coat can serve as a stress raiser in certain circumstances, the trade-off appears worthwhile; i.e., reduced peak intensities outweigh increased lower intensity values. In short our work indicates the gel coat to be a worthwhile concept from the standpoint of load transfer.

To determine why little practical use is made of this concept, the views of Mr. Henry F. Gardner (President, American Academy of Orthotists and Prosthetists) were sought. He believes the principal practical difficulty to be one of migration of the gel under conditions of long time use. That is, the material flows from highly stressed areas to low stress zones, in the course of time, and does not return. We can only note that no permanent set was apparent in the course of our work. It is conceivable that the material, which is viscoelastic, acts differently over long time periods as compared to short period results. It is also possible that the gel material was loaded to higher stress levels in practical socket trials than in the course of our experimental work. Indeed, our load levels were chosen to avoid failure or permanent set in the material; it is likely that these levels were exceeded in the field resulting in failure.

Given the two possible mechanisms of failure stated above, it is our

opinion that the latter (excessive stress) is the more likely cause; we have been unable to cause migration by the long time imposition of moderate stress. As the material concept has merit it may pay to reexamine the process of application. In the view of Dr. Eugene F. Murphy (Director, Research Center for Prosthetics) an improvement may be made through superior anchoring of the gel to prevent lateral displacement. He suggests that the socket interface be given a pattern of ridges to provide restraint. In our view, alteration of the gel material itself is the most profitable road. The gel might be compounded to offer superior stress sustaining properties. The usual price of seeking a higher yield point is greater stiffness, and this is likely to be the case here. In turn the stiffer gel will increase the stress experienced at the stump. Granted this drawback, a revamped gel, stabilized by either superior anchoring or a superior load sustaining ability, appears worthwhile.

Attempts to determine combined stresses along classical lines were not successful. Mohr circle techniques are based on a simple continuum of stress without restraints and certain assumptions such as uniformity of response to tension and compression. The gel material used as a flesh model is subject to numerous constraints and departures from the linear Mohr model. For example, the stress-strain curve employed in our calibration procedure is not a straight line and the Poisson's ratio is clearly different when under tension as compared to compression. Such differences from an ideal material are also displayed by flesh and are therefore desirable. The problem of determining combined stresses for such a difficult material is made awkward by the continually changing "constants," leading to need for an iterative solution, presumably aided by computer techniques. While in principle this may be solved, it is not simple. Ignoring these difficulties and simply solving in the classical manner produces quick answers that are clearly nonsense. The problem of combined stresses has thus proven difficult. We are setting it aside. Clearly there is a challenge for theoretically inclined bioengineers to develop an equivalent of Mohr's circle capable of analyzing complex stresses on various planes of biological tissues. Fortunately, these difficulties do not affect the analytical procedures given earlier in this series of reports. Prior theoretical efforts do not presuppose any fixed relationship between shear and compression or continuity between tension and compression. Therefore, the difficulties encountered in the Mohr circle approach do not apply to earlier work.

Currently, we are testing certain of the concepts developed in the course of this work on amputees with a background of stump problems. The next report will detail this effort.

ACKNOWLEDGMENTS

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