PRESSURES IN CRITICAL REGIONS OF THE BELOW-KNEE PATELLAR-TENDON-BEARING PROSTHESIS [a]

J. Raymond Pearson, M.S.M.E.
Visiting Professor

Prosthetics Technology Research Laboratory
Uppsala University, Sweden

Gunnar Holmgren, C.P.O.
Director of Prosthetic Services

Lennart March
Chief Prosthetist

Kurt Oberg, M.S.A.E.
Research Engineer

Department of Orthopaedic Surgery
Akademik Sjukhuset
Uppsala University, Sweden

INTRODUCTION

In a biomechanical analysis of the forces transmitted to an amputated limb by a below-knee patellar-tendon-bearing prosthesis, Radcliffe (1) showed in 1962 that the principal forces occur in the patellar tendon, the distal anterior tibial, the latera tibial condylar, and the medial tibial condylar regions. Apparantly the design of this prosthesis was such that the forces were directed to those areas in order to transmit them as effectively as possible to the musculo-skeletal system, by a direct route through tissues that could withstand the pressures generated, and to provide continuous control of balance and stability.

In the investigation of prosthesis-limb interface pressures of the above-knee prosthesis in 1967, Appoldt and Bennett (2) found the mean brim load while standing to be roughly 40 percent of the maximum load developed in walking. Similar reductions occurred at other sample sites.

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That investigation also included an exploratory examination of alignment effect. They showed that an abductive alignment increment of 2 deg. from neutral produced small effects in the pressure magnitudes at particular sites, the accumulative effect of which was to shift the time-averaged net loading for stance phase from a posterior to an anterior direction. This suggests an influence on balance and gait. Evaluation of the gait showed a notable change following alteration of the alignment.

Instrumentation and a method of study of such pressures in the below-knee prosthesis were developed by Corell (3) in 1969. By using a grid of five transducers on a ½-in.-sq. area and power-series-curve fits to the data, he was able to show the variation of pressure over such an area and the transmission of a pressure wave over the area as a function of time, as the relative displacement of prosthesis to limb took place. A followup on that investigation, with an additional comparative study of pressure distributions in hard sockets, Kemblo, and gel liners, was reported by Rae and Cockrell (4) in 1971.

A relevant discussion of the techniques and effects of alignment for the above-knee case was presented by Radcliffe in 1954 (5), and the principles and critical effects on function of alignment of below-knee prostheses were reported by Murphy also in 1954 (6). These were also thoroughly discussed by Radcliffe and Foort (7) with regard to the patellar-tendon-bearing below-knee prosthesis in 1961.

The remarkable success of the design of the PTB prosthesis and the minimal number of complaints by patients of pain introduced by alterations of alignment suggest that the pressures encountered are usually tolerable. However, a "graffitus" composed by an amputee that reads "Life is like a prosthesis—always chafing somewhere," curiosity about the effect of alignment on such pressures, and the possible correlation of static pressures encountered in the fitting process with dynamic pressures, lead to this investigation.

OBJECTIVES

Specifically, then, one of the objectives in the study was to measure interface pressures at the critical areas for postural stance positions with the limb suspended, the weight on both limbs, and the weight borne by the prosthetic foot only. It was anticipated that comparison of such values with those for similar configurations of limb and forces under dynamic conditions might lead to a correlation that would be of value to the prosthetist during the process of fitting. A second objective was to measure the effects of alterations of alignment on the critical pressure areas to see where they occur and how severe and sensitive the pressure changes are.

THEORETICAL PRESSURE ESTIMATES

As a point of reference and basis of judgment of the pressures to be measured, estimates of a conceptual uniform normal pressure were made. The first of these assumed total weight borne by the prosthesis over the projected surface area of the stump, which is equal to the maximum cross-sectional area of the stump. This includes an assumption of end-bearing with equal distribution of pressure over sides and end. The following is the evolution of an expression for such a uniform normal pressure distribution.

$$p_{una} = \frac{W}{\pi D_x^2/4} = \frac{4\pi W}{(\pi D_x)^2} = \frac{4\pi W}{C_x^2} \qquad (1)$$

with pressure p, weight W, diameter D, and maximum periphery C_x.

With no end-bearing this expression modifies to

$$p_{unb} = \frac{4W}{\pi(D_x^2 - D_n^2)} = \frac{4\pi W}{C_x^2 - C_n^2} \qquad (2)$$

It might be supposed that a uniform pressure value would usually fall between that given by these two expressions. It should be noted, however, that neither of these accounts for the effect of friction which could be significant with no sock and no perspiration lubrication.

EXPERIMENTAL METHODOLOGY

Fitting Procedure

The neutral position of alignment was established by the normal procedures of the prosthetist during fitting. His judgment of a properly aligned prosthesis for the subject at hand was taken as the neutral setting. The procedure as described below is based upon recommendations of the PTB below-knee manual (7) and practices developed by the prosthetist authors of this account.

An artificial foot of the size and type appropriate for the case in point is inserted into the shoe to be used by the amputee. The top plane of the artificial foot or ankle block is then set in a horizontal plane. The orientation of the plane is checked with a level and altered, if necessary, by modification of the heel height, or the dorsiflexion stop with ankle-jointed feet, until the plane is level. A plumb line is then used to see if the axis of the pylon is vertical. The socket-shell-support plate of the alignment coupling is then set level in a horizontal plane.

When no indication exists for special treatment of a varus or valgus condition, the routine procedure for prealigning the socket in space

is to mount the prosthesis socket shell on the alignment-coupling-support plate with its longitudinal axis rotated 5 deg. anteriorly and 5 deg. laterally about axes at the center of the support plate. The inclination of the socket axis may be fixed by cutting the bottom surface of the socket-shell base block, when the top surface is inclined, so as to give the socket axis the inclination described above. The socket axis can also be oriented by setting the socket shell in the base block with the orientation specified. In this case both top and bottom surfaces of the base block will be horizontal. The socket is thus prealigned with an orientation in space that induces 5 deg. of knee flexion, so as to optimize extension of the knee up to, but not more than, 175 deg. This prevents hyperextension and knee lock, assuring continuous control of knee flexion. The prealignment also induces 5 deg. of shank abduction for concentration of pressures in the medial tibial condylar region.

The prosthesis is now fitted to the stump and post-fit alignment by coupling adjustment can take place. The horizontal position of the pylon and foot with respect to the socket axis is set by the anterior-posterior and lateral-medial adjustments of the alignment coupling. Those positions are checked by plumb line with the procedure described below.

Alignment in the parasagittal plane is checked by setting the origin of the plumb line at the projection of the center of the greater trochanter. The line should intersect the lateral-medial axis of the knee joint and line up with the anterior element of the pylon. The projection of the lateral-medial joint axis is taken at the mid-point of the tibiofibular joint or at a point midway between the anterior and posterior brims of the socket shell.

In the frontal plane the ischial tuberosity, the knee center, and the axis of the pylon are aligned and checked by plumb line with origin at the ischial tuberosity. A point at two-thirds of the distance from lateral-to-medial surfaces of the thigh on a line at the ischial tuberosity level is taken as a landmark for the tuberosity. The projection of the anterior-posterior axis of the knee occurs at a point 0.5 to 1.0 cm. medial to the socket axis at the brim level.

The amputee is now observed from behind and in front as he walks to ascertain that the plane of motion of the prosthetic leg during swing phase is parasagittal, and that symmetry of motion of the two limbs exists.

Certain techniques of observation are used to relate abnormalities of gait to definitive misalignments for which corrections are made with the coupling. Also, the prealignment procedure may be altered to account for abnormal orientation of the thigh or shank brought about by contractures or other causes.

Alignment Coupling

The VAPC below-knee adjustable shank alignment coupling, as fabricated by the United States Manufacturing Company, was used for alterations of alignment adjustment. As may be seen from Figure 1, this coupling permits anterior-posterior and lateral-medial adjustments of 10 mm. by virtue of the dovetail guides. Alteration of the positions of the tilt-plate wedge nuts while maintaining the distance between them, i.e., tightening one and loosening the opposite, provides an angular adjustment of up to 10 deg. This coupling has a more limited range than others considered, but it has the valuable characteristic of permitting any one of the adjustments to be made without disturbing the others.

Transducers and Locations

Pressure measurements were made with a transducer developed especially for this use by the Kulite Semiconductor Products Corporation in collaboration with the University of Michigan Orthotics Research Project. The development and characteristics of the device are described by Corell, Rae, and Cockrell (3 and 4). The transducer is 0.318 cm. (0.125 in.) in diameter and 0.076 cm. (0.030 in.) in thickness. It is not felt by the subject when inserted between the prosthesis shell and the limb.

The transducers were located in the patellar tendon, distal anterior tibia, lateral tibial condyle, and medial tibial condyle regions. The

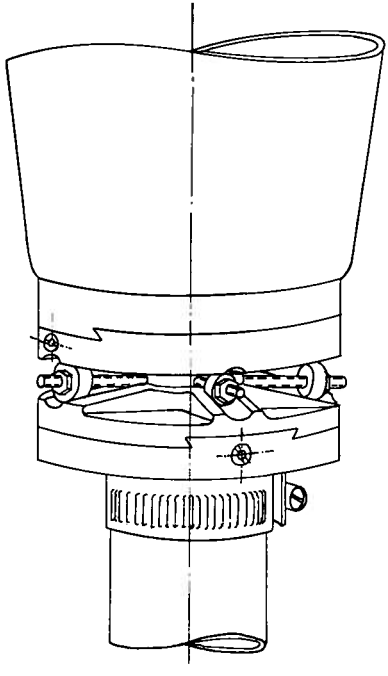

FIGURE 1.—Sketch of VAPC below-knee adjustable shank alignment coupling.

sensitive diaphragm of the transducer was placed against the skin and the device taped in place. The patellar-tendon transducer was located midway between the lower edge of the patella and the tibial tubercle. In the distal anterior tibial region, the transducer was located just above the edge of the bevel of the amputated tibia on the tibial flare and near the crest. The condylar transducers were placed on the flare leading up to the condyle in predominantly lateral and medial positions to sense frontal-plane force effects.

Goniometers and Foot Switches

In order to correlate pressure variations with limb configurations and foot positions during the walking cycle, simple flexible strip contactors were installed at the heel and toe of the shoe. The subject walked on a strip of paper with heel and toe inked to provide footprints from which step measurements could be made. A coordinator switch produced a "blip" on the recorder trace that was associated with a given step on the footprint record. Goniometers were fabricated for the knee and hip, consisting of bars attached to adjacent segments of the limb with Velcro straps, and simple wire-wound element resistance potentiometers were located at the joint centers. A circuit box containing a power source and a balance potentiometer was attached to one of the bars. Spacers accounted for limb shape variation and prevention of binding. Very light, flexible, 2-mm. leads from the foot-switch junction box, potentiometer circuit boxes, and pressure-transducer meter and signal-conditioner box, brought the output to an Elema-Schönander Mingograph (sic) 81 ink-jet oscillograph recorder. The leads were about 10 meters long, sufficient to easily permit three patterns of five strides each with recorder located at the walkway midposition.

Test Regime

After gathering pertinent information with regard to the subject's weight, height, stump measurements, and locations for the transducers, the latter were attached and their output readings zeroed. The prosthesis was then put on and the subject was requested to follow the regime listed below:

REGIME

1. Subject standing with sound foot on a platform and prosthesis suspended. Stump muscles relaxed and then tensed.
2. Subject standing with weight on both feet. Stump muscles relaxed and then tensed.
3. Subject standing with weight on prosthetic foot only. Stump muscles relaxed and then tensed.

4. Pylon set 10 mm. laterally from the neutral position. Posterior-anterior alignment adjustment maintained at neutral.
5. Subject walks five strides at his normal cadence. (This provided three "steady state" strides, bracketed by one of acceleration and one of deceleration.)
6. Pylon set at 5 mm. lateral from neutral. Subject walks five strides.
7.–9. This process was continued until the lateral-medial adjustment range had been scanned.
10. Coupling reset to lateral-medial neutral position.
11.–15. A similar procedure was followed in scanning the posterior-anterior adjustment range.

A similar regime was followed in testing for angular alignment adjustments on two of the 10 subjects. The manner of setting the angular alignment adjustments is described below under Results.

Subjects

A tabulation of the subjects and their measurements appears in Table 1. They were all males, ranging in age from 31 to 76, in sound general health with well stabilized stumps. Subject 1 was a vigorous athlete. C_x is an abbreviation for the *maximum* perimeter of the stump measured at the level of the tibial tubercle. C_n is the *minimum* circumference measured at the level of the end of the amputated tibia. H represents the axial distance between those two circumferential measurements.

TABLE 1.—*Subject Measurements*

No.	Initials	Age yrs.	Weight kg.	Stump dimensions, cm.			Amputation age, yrs.
				C_x	C_n	H	
1.	K.M.	31	74.0	31.0	25.0	11.0	6.5
2.	S.N.	37	67.0	30.0	22.0	11.0	5.0
3.	F.M.	45	85.0	32.0	22.5	15.0	19.5
4.	A.A.	56	90.2	37.0	24.5	13.5	2.5
5.	A.W.	47	78.2	32.0	20.0	10.0	23.0
6.	T.S.	76	68.0	26.0	21.0	15.0	58.0
7.	H.A.	58	96.9	34.0	20.0	8.5	9.5
8.	H.H.	64	77.3	32.5	29.0	12.5	1.0
9.	S.F.	53	64.7	31.0	21.0	14.0	27.0
10.	J.S.	65	89.7	39.0	18.0	11.5	42.0

RESULTS

Uniform Normal Pressure Estimates

Values for the uniform normal pressure estimates were computed with the expressions developed above and appear in Table 2. The column

under "a" represents the usual "rule of thumb" for such pressure estimates. It might be noted that neither of the columns includes consideration of the effect of friction. A comparison of the two columns reflects the effect of total end-bearing, indicating that the side pressures are about double for no end-bearing. The standard deviation is given simply to give an impression of the degree of variation for this set of subjects. Inference that this value would hold for a large population of such subjects is not intended. The coefficient of variation is the ratio of standard deviation to average in percent and is simply another way of expressing the variation encountered. As these values were computed for total weight, they will be compared with the cases of weight borne by prosthetic foot alone and mid-stance pressures of the walking cycle.

TABLE 2.—*Estimates of Uniform Normal Pressure*

No.	Initials	Uniform normal pressure, kg/cm²	
		a.	b.
1.	K.M.	.97	2.08
2.	S.N.	.94	2.02
3.	F.M.	1.09	2.28
4.	A.A.	.83	1.48
5.	A.W.	.96	1.58
6.	T.S.	1.27	3.16
7.	H.A.	1.05	1.61
8.	H.H.	.92	4.15
9.	S.F.	.85	1.46
10.	J.S.	.76	.96
	Averages	.96	2.12
	Std. Deviation	.14	.89
	Coef. of Variation	14.5%	42%

Postural Stance Pressures

The pressures in the regions examined for three modes of support in postural stance, namely, prosthesis suspended, double support, and prosthesis support only with stump muscles relaxed, are shown in Figure 2. Comparison of the values for any given site show the effect of transferring weight onto the prosthesis. Comparison of the same column a, b, or c for the four groups indicates the manner in which the pressures are distributed among the various regions. The uniform normal pressure estimate indicated on the prosthesis support column provides an idea of the degree of concentration of pressures on the critical areas and clearly indicates that the pressures are not uniform, even though these were total-contact fittings.

The crosses appearing in the columns are the standard deviations from the averages for the 10 subjects. These deviations are corrected for small samples, but, as before, they are there simply to indicate the order of magnitude of the variation. They are plotted from the average value, either up or down, for convenience of plotting without regard for sign.

Of note here is the fact that the pressures in the distal anterior tibial region are of the same order of magnitude as those of the patellar tendon. As the amount of soft tissue tends to be less in the distal anterior tibial region, it appears to be particularly critical.

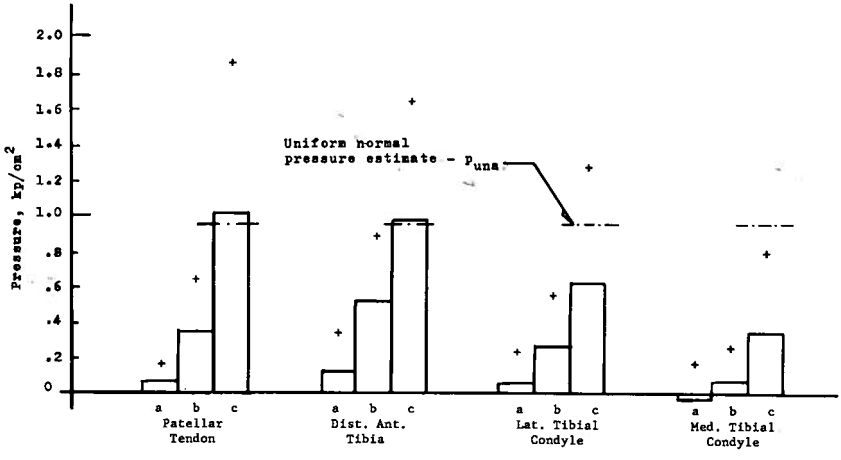

FIGURE 2.—Average values of the pressures in the critical regions for three modes of support in postural stance, i.e., prosthesis suspended—a, double support with weight on both feet—b, and prosthesis support only—c. Stump muscles are relaxed. Points indicate standard deviations from average.

During the process of testing, it was noted that tensing of the stump muscles could have a significant effect upon the interface pressure. Hence, some of the subjects were requested to tense the stump muscles after the values for the relaxed condition were taken. The results of these readings are plotted on Figure 3 where the distribution is similar in nature, but the pressure values are larger for all cases. The averages of these values are for six of the subjects. From this comparison, one may deduce that muscle tension can produce values of the order of magnitude of *twice* those that occur when the muscles are relaxed, and that they exceed the uniform normal pressure estimate radically in the patellar tendon and distal anterior tibial regions. The latter is to be expected, of course, since the muscle contraction effect was not included in deriving the expression for the estimate.

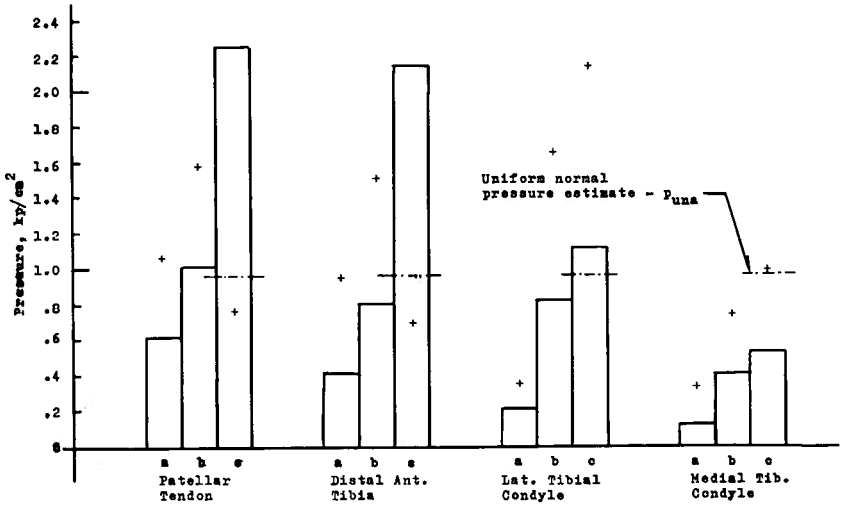

FIGURE 3.—Average values of pressure in the critical regions for three modes of support in postural stance, i.e., prosthesis suspended—a, double support with weight on both feet—b, and prosthesis support only—c. Stump muscles are tensed. Points indicate standard deviations from average of six subjects.

Comparison of Maximum Dynamic and Postural Stance Pressures

In order to correlate static and dynamic pressure values in search of a criteria test, the maximum pressures that occurred during the walking cycle are plotted in Figure 4 for comparison with the two previous plots. The columns designated b and c are those for postural stance, double support, and prosthesis support, respectively, for the case of tensed muscles as found in Figure 3. The third columns designated d are the average values for the 10 subjects of the maximum pressures that occurred during the walking cycle near heel-contact or push-off and with alignment set at neutral. Comparison of the d columns with the other two provide an impression of the effects of the dynamics of the walking cycle. The comparison suggests that these so-called dynamic pressures include the muscle contraction effect as well as those of inertia and the leverage effect brought about by the change in limb configuration.

Extreme Alignment Effects

The effect of the maximum posterior alignment of the pylon and foot can be seen in Figure 5, where the pressure variations of stance phase for neutral and maximum posterior alignment are compared. All figures are drawn to the scales indicated for the "steady state" condition for the lateral condyle region. The diagrams are arranged in three columns, the

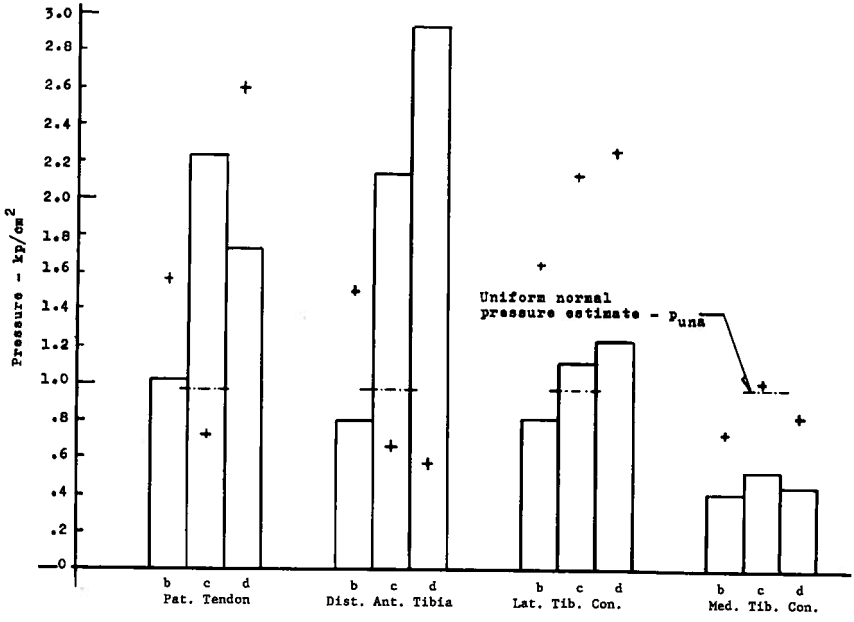

FIGURE 4.—Average values of pressures with tensed stump muscles in postural stance double support—b, prosthetic support only—c, and maximum dynamic pressures occurring at heel-contact.

first of which is that of the initial step, followed by one of the so-called "steady state," and the final step. The cases were separated out in this fashion to distinguish between situations where acceleration of the first and deceleration of the last steps were apt to predominate in influence. As indicated before, each test was of a five-step pattern in an effort to produce the "steady state" of normal cadence. These curves are those of a single subject and are presented here as a characteristic case.

Examination of the "steady state" column indicates that alterations of alignment affected the phasing or time relationship of the pressures. Peak values occurred later for maximum posterior alignment than for neutral alignment. The lateral condyle-initial step curve is a case in which the magnitude of the peak value is increased with little time shift, and the lateral condyle—final step case is one in which magnitude is increased and the peak is delayed as well.

In subsequent examination of maximum dynamic pressure values, those of the first peak following heel-contact or of the second peak at push-off were selected for comparisons, since the larger of the peak values is likely to be of greatest concern in the problem of fitting. These values were taken from the "steady state" steps, although the peaks of start and stop were often larger. It should be noted that higher values can occur in turning, ramp or stair climbing, or other such maneuvers.

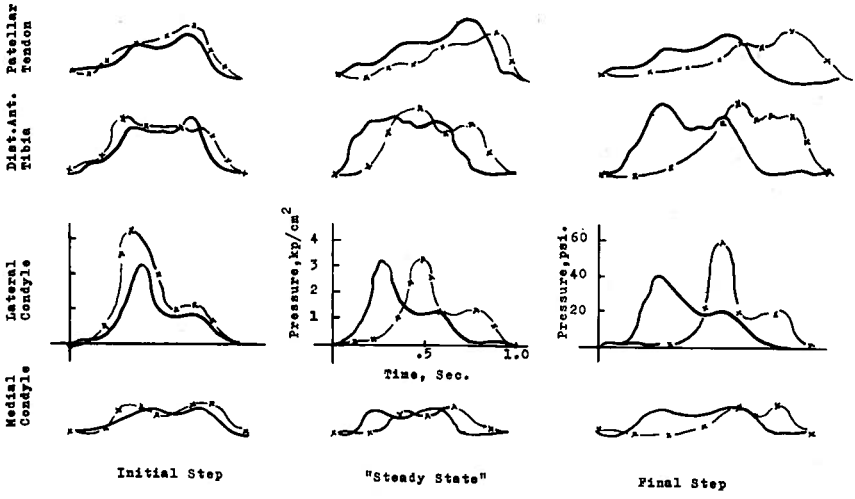

FIGURE 5.—Comparison of cycles of pressure variation of the critical regions for neutral (.....) and maximum posterior (.....x.....x.....) alignment offset for subject 1.

For this case in particular, adjustment of alignment from neutral to maximum posterior had the effect of delaying peak values, but had little effect upon their magnitudes.

Figure 6 is a similar set of curves in which the pressure values for neutral alignment are compared with those of maximum anterior align-

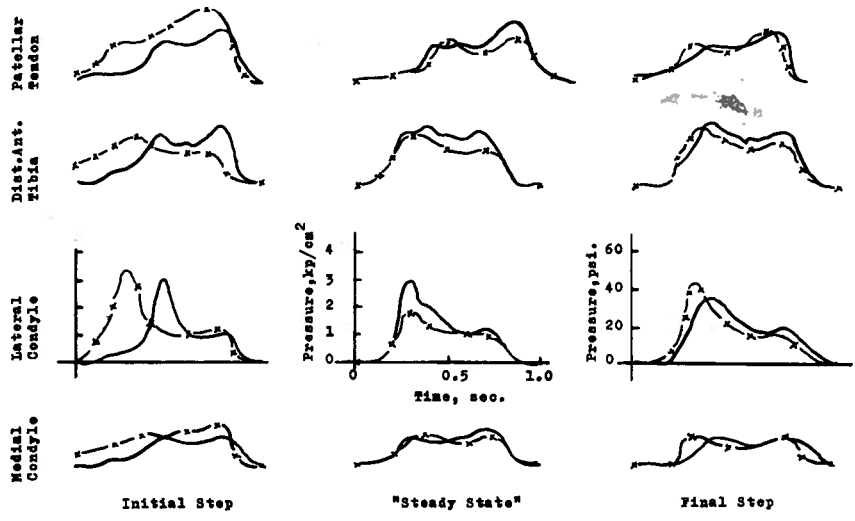

FIGURE 6.—Comparison of cycles of pressure variation for the critical regions for neutral (.....) and maximum anterior (.....x.....x.....) alignment offset for subject 1.

ment. The curves are for the same subject, and so the neutral alignment curves are the same as those of the previous figure. The initial and final steps indicate a timing advance of the peaks. There is a temptation to relate these pressure effects to those expected for force and moment changes due to alignment alterations, but it must be borne in mind that the amputee can compensate for the change by any of several alterations of limb configuration as he adapts to the different alignment. Also, the modification affects the position of the prosthesis shell with respect to the limb, which affects the pressure distribution in turn.

Lateral-Medial Offset

The effect of lateral offset can be seen in Figure 7, where the pressure variations for the neutral alignment case are compared with those of maximum lateral offset. There is a slight difference in the stride period for these two cases, which is evident in the heel-switch record. The maximum pressure at the patellar tendon showed an increase of about 50 percent and the lateral condyle pressure more than doubled, while values for the medial condyle showed little change beyond some phase shift caused by the stride period difference. The large increase in lateral condyle pressure suggests a change in the frontal plane knee moment, such as to increase the reactions in the proximal-lateral and distal-medial regions.

Similar curves comparing the pressure variations of neutral alignment with those of maximum medial offset are seen in Figure 8. Here the stride periods are the same and heel-contact times are matched, but hip and knee actions appear to be delayed. The phase shift is reflected in the pressure curves, since the pressure variations are dependent upon the joint actions. If the swing-phase knee-flex peaks are matched, the pressure curves appear to be similar, indicating a minimal effect of medial-alignment adjustment for this subject.

These curves and those for posterior-anterior alignment adjustments indicate that the pressure peak at push-off is sometimes larger than that of heel-contact.

Effect of Posterior-Anterior Alignment Adjustment on Average of Maximum Dynamic Pressures

From data of the kind seen in Figures 7 and 8, the maximum dynamic-pressure averages for eight of the 10 subjects and for each of the four regions were collected and plotted. The values were taken from one of the so-called "steady state" cycles, with an effort to match stride periods to minimize the effects of changes in cadence and consequent accelerations and inertial forces. Pressure readings were taken for at least five alignment settings ranging from a posterior setting of 10 mm. (approximately

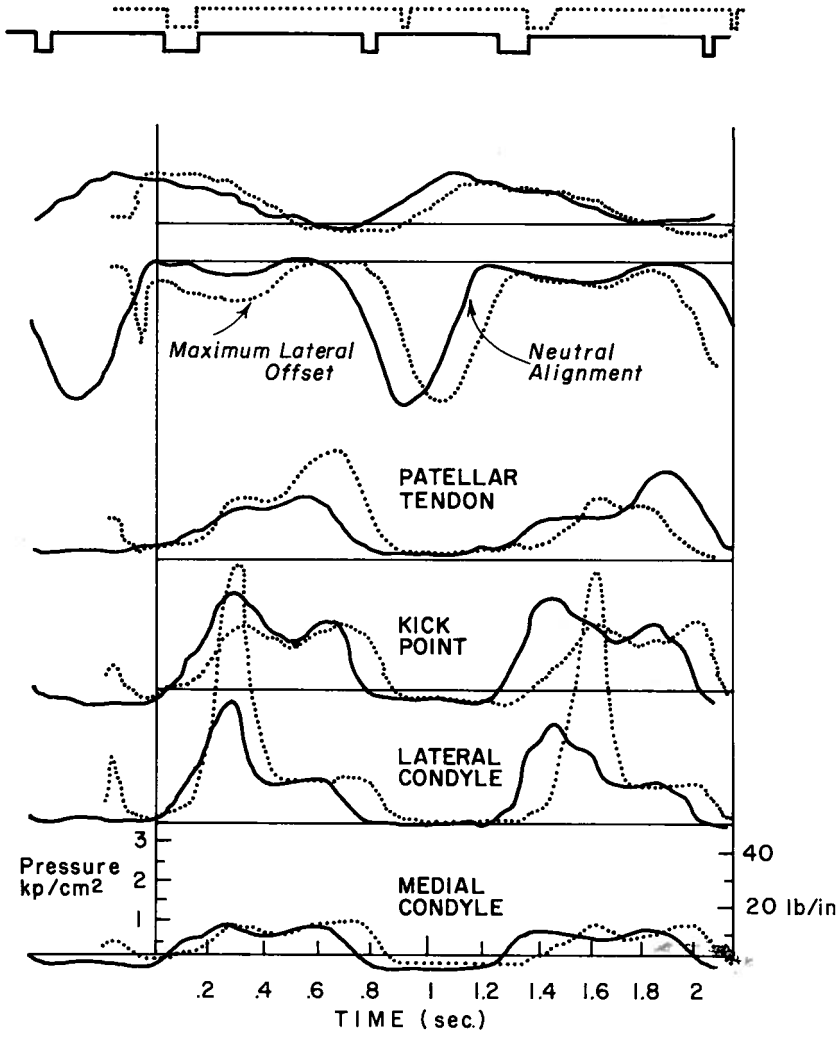

FIGURE 7.—Comparison of pressure variations of neutral alignment and maximum lateral offset for subject 1. Full traces are for neutral, dotted for maximum lateral offset.

⅜ in.) to an anterior setting of 10 mm. This was the maximum range permitted by the coupling. The values were plotted on Figure 9. The straight line curves are least-square fits to the data that indicate a trend for the subjects examined. As the coefficient of variation for the average values of pressure is of the order of magnitude of 60 percent, these curves cannot be regarded as precise indicators of the correlation of these two variables for the general population. However they can be used to draw

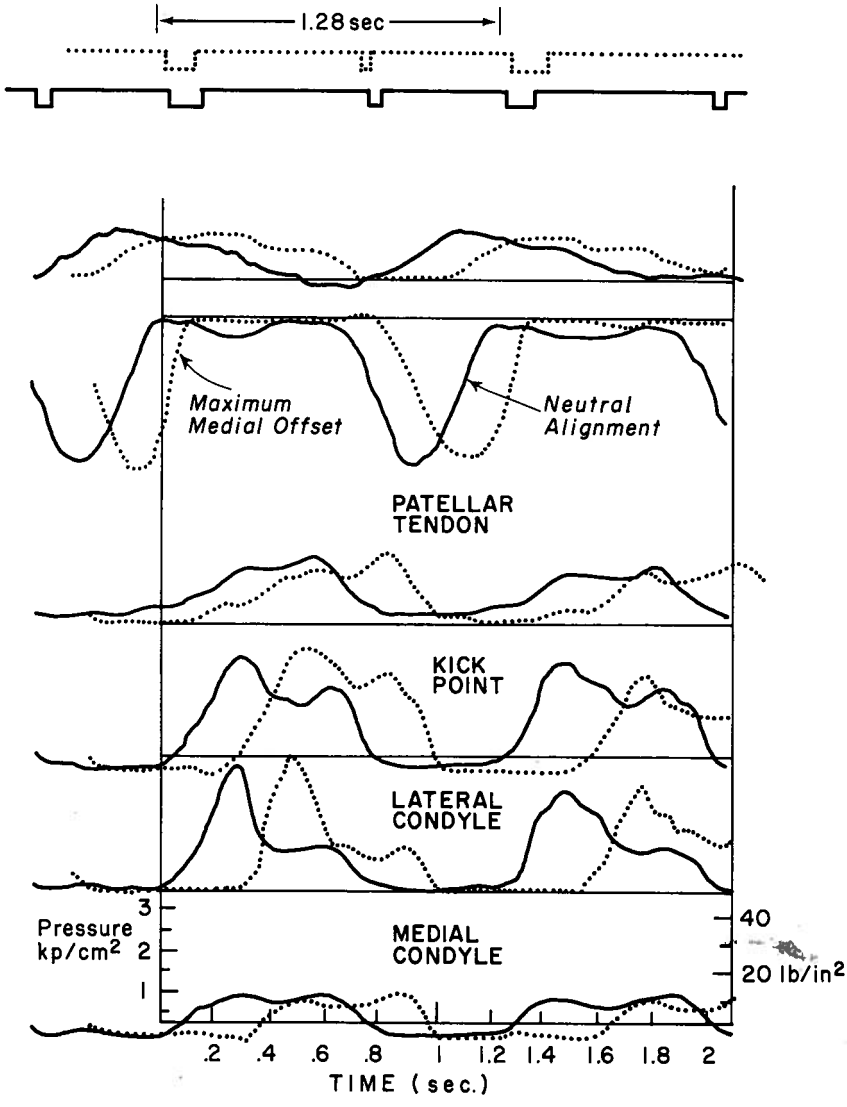

FIGURE 8.—Comparison of pressure variations of neutral alignment and maximum medial offset for subject 1. Full traces are for neutral, dotted for the maximum medial offset.

some conclusions about certain trends and can be tested with analytical reasoning.

The first thing that might be noted is that the pressures for the distal anterior tibial region exceeded that of the other regions, and that the pressure there decreases as alignment is adjusted from posterior to

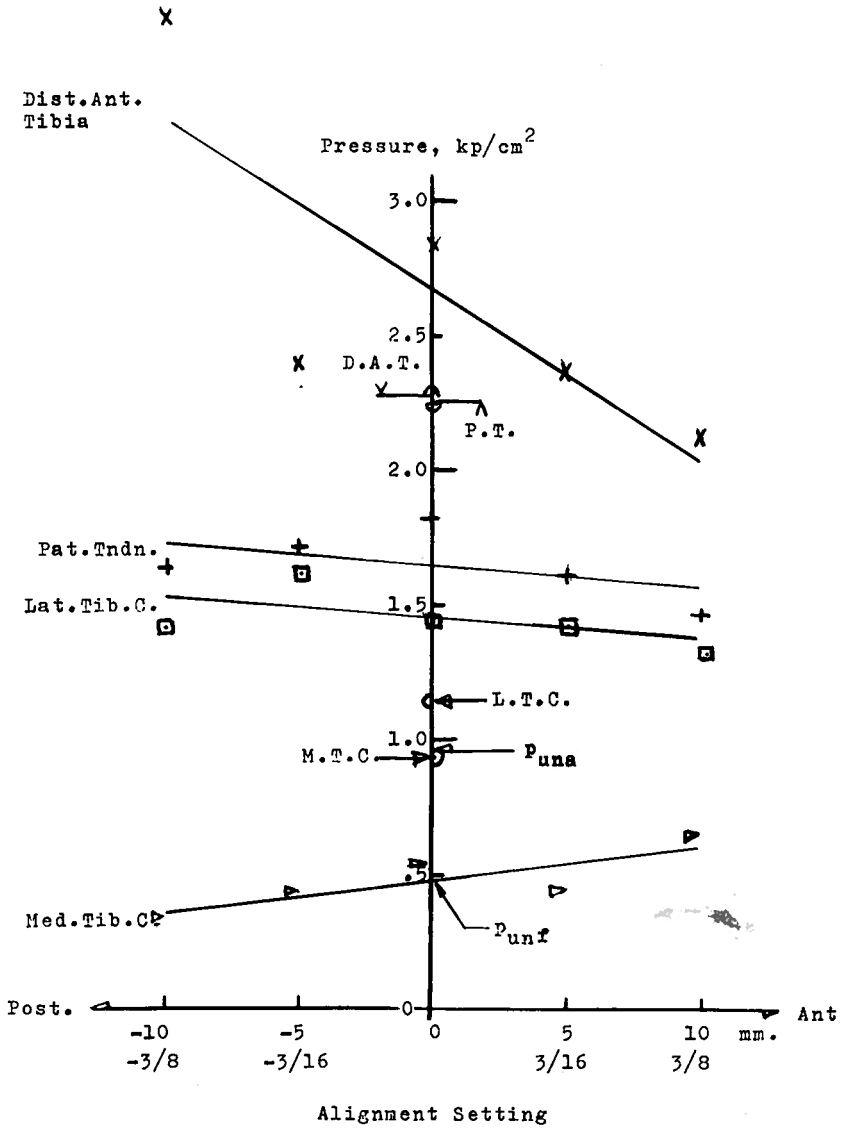

FIGURE 9.—Average maximum dynamic pressures occurring at the patellar tendon (+), distal anterior tibia (×), lateral tibial condyle (□), and medial tibial condyle (Δ) regions for varying posterior-anterior alignment settings. Semicircles on the ordinate indicate average static pressures for total weight borne by the prosthesis with stump muscles tensed. Estimated uniform normal pressure, P_{una}, is also indicated.

anterior. Furthermore, the slope of this line is greater than any of the others, indicating that this region was more sensitive in pressure changes to alterations of alignment than the others. Since such alignment alterations in the sagittal plane reduce the knee moment, and the force at this point is a reaction to that moment, the trend seems to be likely. As soft-tissue thickness is minimal in that region, pressure tolerance is relatively low, and so the effect is an important one.

The pressures in the patellar tendon and lateral tibial condyle regions are similar and show little change of pressure with alignment alterations. The medial tibial condyle region showed the lowest pressures, although the preliminary alignment of the socket in space and the design concept of the PTB prosthesis would suggest that predominant forces should occur at the patellar tendon and in the medial condyle region. Because of the bony protuberances here, it is likely that the distribution of pressure is radical and that placement of the transducer here is critical. This region does seem to be somewhat more sensitive to alignment changes, and the pressure appears to increase with alignment alterations in the anterior direction.

The localized static-pressure levels with stump muscles contracted and weight borne on the prosthesis are indicated by the semicircles on the ordinate. Though not precise indicators of the maximum dynamic-pressure levels, they do appear to be of similar order of magnitude. It is obvious that the uniform normal pressure estimate would not serve this purpose as well. The estimate p_{una} is the "rule-of-thumb" estimate with full end-bearing assumed. The other estimate, p_{unf}, is for partial end-bearing and a frictional effect included.

Effect of Lateral-Medial Alignment Adjustment on Average Maximum Dynamic Pressures

A similar set of curves demonstrating the effect of lateral-medial alignment adjustments occurs in Figure 10. The sensitivity of the distal anterior tibial region to alignment changes is much less than that of posterior-anterior adjustments and is opposite in sign; i.e., the pressure increases slightly with anterior adjustment. One might have anticipated more radical effects in the condyle regions, but it seems not to have been the case for these subjects. The relation of the static-pressure levels and the estimates are just about the same as in the previous case.

Effect of Flexion-Extension Angular Alignment Adjustments on Maximum Dynamic Pressures

Angular alignment adjustments were managed by the use of three fixed-angle gages with plates set at 0, 5, and 10 deg. One plate was set

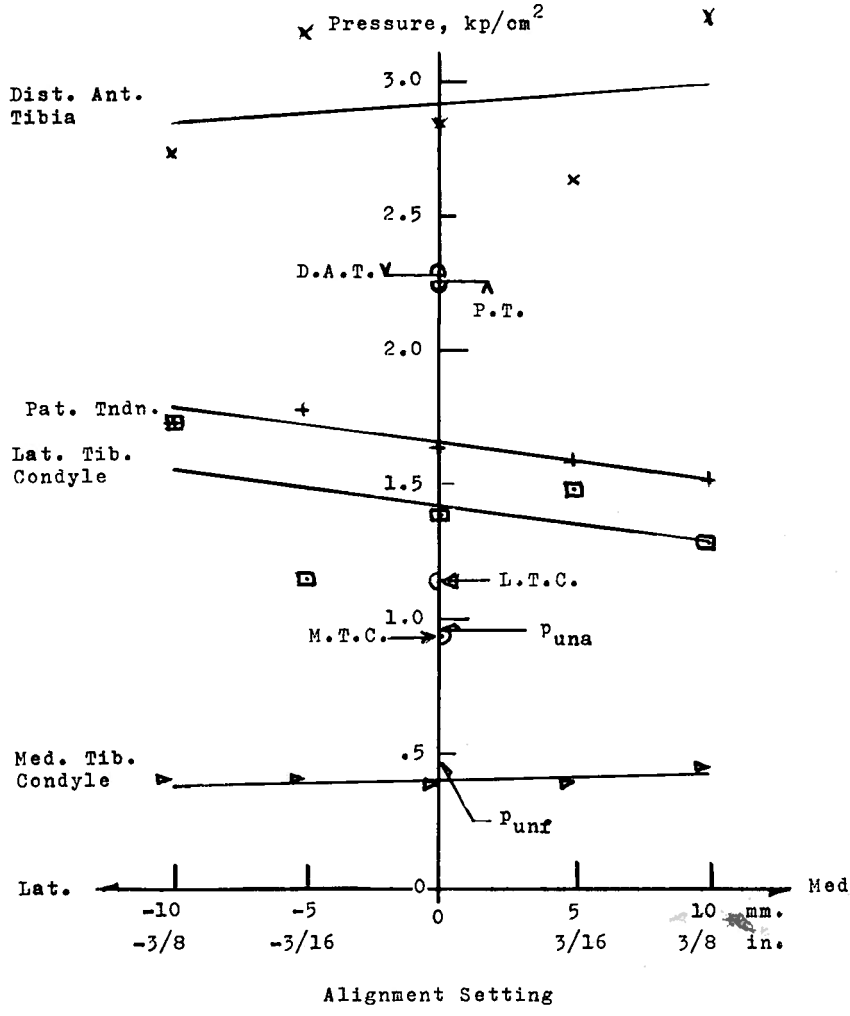

FIGURE 10.—Average maximum dynamic pressures occurring at the patellar tendon (+), distal anterior tibia (×), lateral tibial condyle (□), and medial tibial condyle (Δ) regions for varying lateral-medial alignment settings. Semicircles on the ordinate indicate average static pressures for total weight borne by the prosthesis with stump muscles tensed. Estimated uniform normal pressure, P_{una}, is also indicated.

flush with the prosthesis shell-block bottom, and the other with the bottom surface of the coupling plate which was attached to the top of the pylon. The plates had semi-circular openings that permitted them to fit around the upper coupling plate and the pylon. These angle gages provided five alignment adjustments from one extreme to the other.

The data for the results of this test on one subject are presented in

Figure 11, where the maximum dynamic-pressure averages occurring in the critical regions for alignment-adjustment increments from extreme flexion to extreme extension about the coupling axis may be seen. It should be borne in mind that angular-alignment modifications result in much larger displacements of the foot than do the simple linear adjustments examined above. For a typical subject with a pylon-foot length of 33 cm., a 10 deg. sagittal-plane angular-alignment adjustment results in a foot displacement of 57.5 mm. This is close to six times greater than the maximum linear adjustment for the coupling used. Hence these curves reflect the effect of large foot displacements. This effect is detectable in the steep slope of the distal anterior tibial curve, reflecting greater sensitivity than in foregoing cases.

It should be noted also that such angular adjustments change the angle of the foot radically. In the normal postural position or at midstance when the knee is in full extension, for example, a flexural angular adjustment will lift the heel, transferring the point of application of the floor reaction to the front of the foot even though translation of the foot is backwards. Such a movement of the floor reaction will certainly reduce the knee moment and might even change its sign, so that the predominant force will be shifted from the distal anterior region to the patellar tendon. In other words, it can result in shifting from a flexion to an extension moment applied to the stump by the ground reaction.

Conditions at heel-contact and at push-off are somewhat more complex, since this type of adjustment alters the point of contact with the floor. Contact is reestablished by adaptive changes in the configuration of the limb system, and force changes will depend upon how adapatations are made. As there are several different ways in which this can be done, it is difficult to foretell the effect without knowing which mode of correction the subject uses.

Effect of Abduction-Adduction Angular Alignment Adjustments on Maximum Dynamic Pressures

In the abduction-addition alignment adjustments, the effect on the patellar tendon appeared to be similar though generally lower than the foregoing case, as may be seen in Figure 12. Pressures on the distal anterior tibia are still high and sensitive though the slope of the curve has changed, indicating a decrease in this pressure as the pylon is adducted. The decrease in pressure and increase in sensitivity for the lateral tibial condyle region is as one might expect from a medial movement of the foot. As is generally the case, the medial tibial condyle region shows little effect of the alignment alterations.

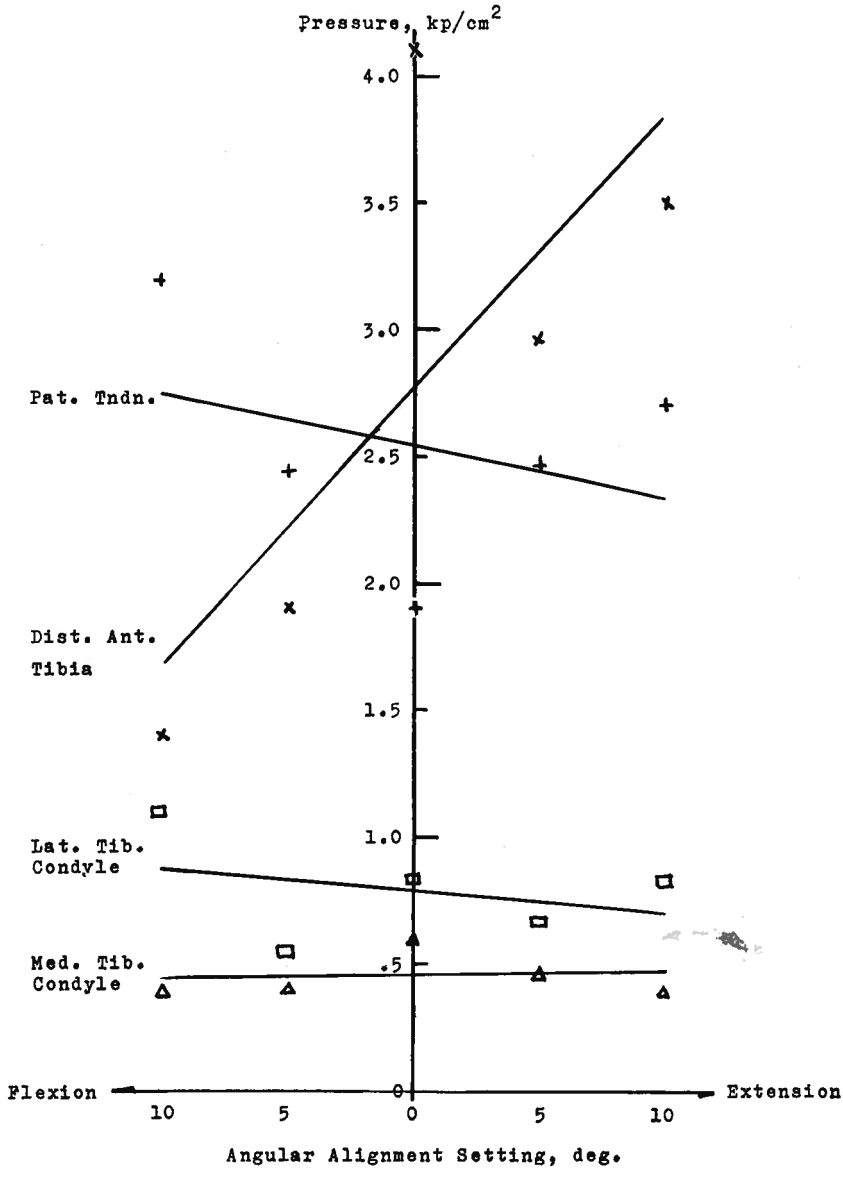

FIGURE 11.—Maximum dynamic pressures occurring at the patellar tendon (+), distal anterior tibia (×), lateral tibial condyle (□), and medial tibial condyle (Δ) regions for varying angular alignment settings for subject 7, from extreme flexion to extreme extension of pylon and foot about the coupling axis.

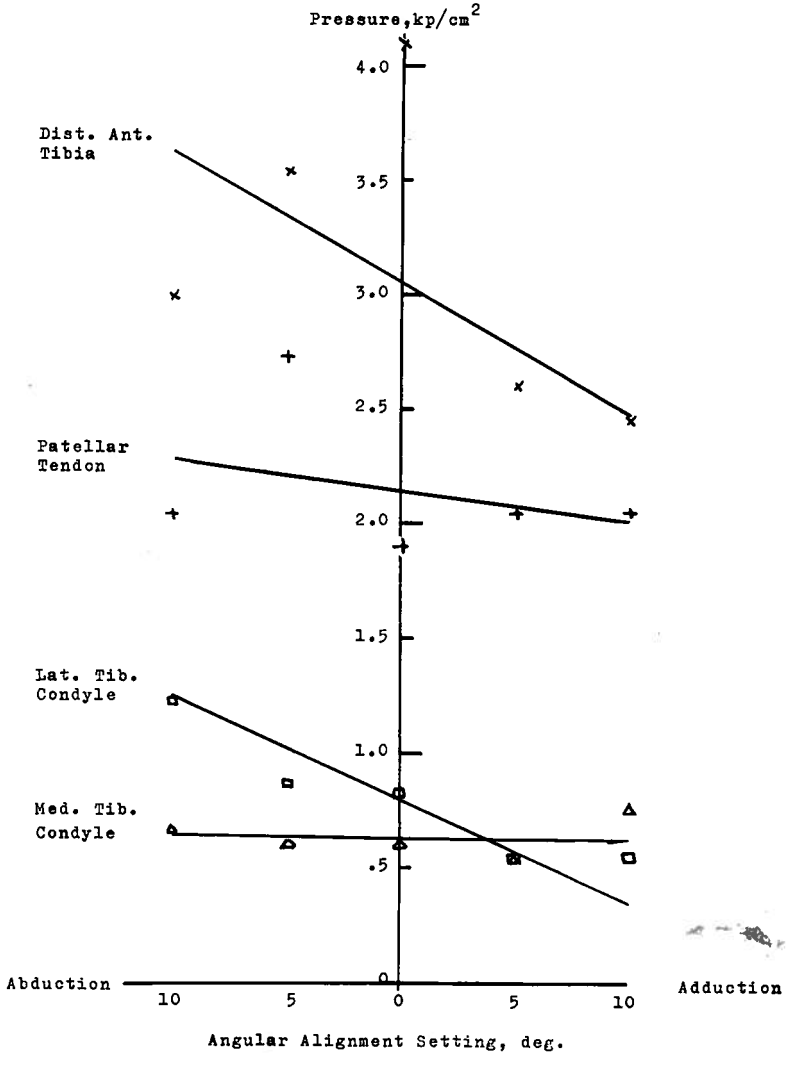

FIGURE 12.—Maximum dynamic pressures occurring at the patellar tendon (+), distal anterior tibia (×), lateral tibial condyle (□), and medial tibial condyle (△) regions for angular alignment settings varying from extremes of abduction to adduction for subject 7. Rotation is about the coupling axis.

DISCUSSION

Pressure Factors

A consideration of the foregoing results and the comments pertaining to them indicates that there are several factors that affect these interface pressures. Variation in pressures recorded for the suspended prosthesis

from one subject to another is an indication that the original fit can alter the pressure. It seems likely that the tension in the bandage that is used to wrap the stump in forming the plaster negative mold influences this prime pressure. Even though that process produces total contact at the interface of limb and prosthesis shell, the pressures will vary as the forces are transmitted through tissue of varying stiffness to bone and thence into the musculo-skeletal system. Still, pressures of original fit are relatively small compared to those caused by force or weight transmission, as may be seen by comparison of the different modes of support in Figure 2. These are exceeded in turn by the effect of muscle contraction, the radical effect of which was noted in Figure 3. Another significant increment of pressure is added by the inertial and leverage effects encountered in the walking cycle, as seen in Figure 4.

Cadence

The influence of cadence on these particular measurements was minimized by having the subjects walk at their "normal" spontaneous cadence. Even that condition, however, would result in some variation, though it was controlled in individual subjects by selecting pressure values from cycles showing minimal change from adjacent cycles. An impression of its degree of influence can be reached by the following reasoning: The predominant pressures always occurred at the distal anterior tibia and the patellar tendon. The forces that produce these pressures are highly dependent upon the knee moment in stance phase, the maximum of which occurs at or near the maximum knee-flexion angle (8). A plot of this angle as a function of cadence from data extracted from Reference 9 appears in Figure 13. The cadences for the amputee subjects of this investigation were in the low range of approximately 60 to 90 steps per minute where the changes in stance-phase knee-flexion angle are smallest. The minimal slope of the curve in this range bears out this point. It can be concluded then, that the cadence differences had little effect upon the pressure values used for particular subjects, but probably did have an effect from subject to subject and therefore influenced the averages. It is, of course, but one of several factors that influenced the variation in pressure.

Transducer Location

Effort was made to locate the transducers at a point where the maximum pressure for that region was expected. Nevertheless, there was some uncertainty as to whether the values were the maximums or some proportion thereof. In dynamic measurements, the pressure is a function of time as well as location. There is a pressure wave that crosses the

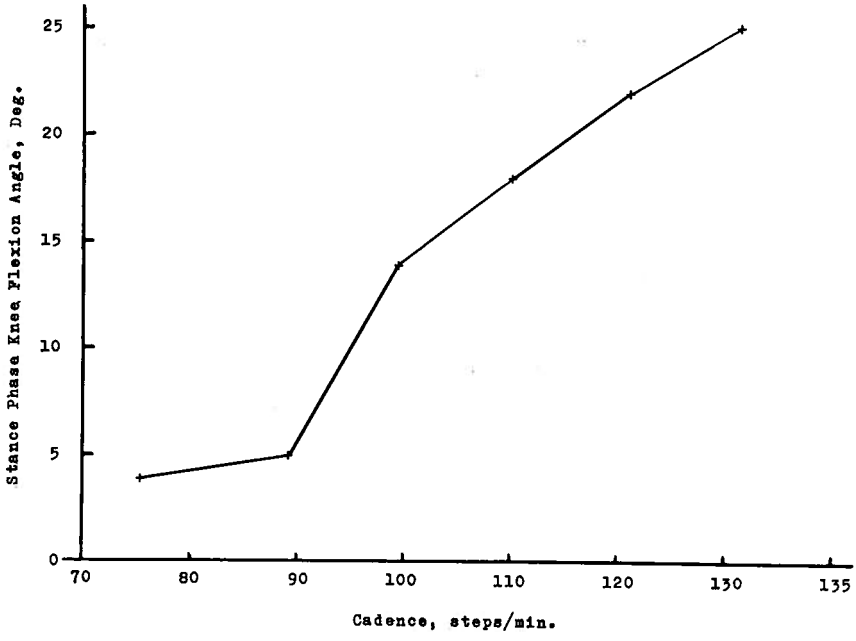

FIGURE 13.—Stance-phase knee flexion as a function of cadence. Data from Figure 66 of Reference 9.

area in question and creates a scanning effect. The scanning effect compensates, at least for minor misplacements (3).

Correlation of Static and Dynamic Pressures

One of the original objectives of this investigation was to seek a correlation between static pressures of fit and dynamic pressures of walking that might be of use to the prosthetist. In this respect, examination of Figure 4 suggests that for prosthesis support only with stump muscles contracted, the pressures are at least rough approximations of the maximum dynamic-pressure averages. Such static pressures might then be an indication of what to expect. The dynamic-pressure values of Figure 4 are those for neutral alignment and do not therefore predict the effects of alignment. In most cases, however, these are small as compared to the error of approximation.

The value of these pressures as a criterion of judgment of the quality of fit and alignment depends upon a definition of tolerance levels. The pressures are rarely sufficient to cause pain. They often exceed the local peripheral blood vessel pressures, but their cyclic quality turns this into a beneficial pumping effect that promotes tissue health. It seems likely that such tolerance levels will eventually be defined in terms of

the cumulative effect of pressure variations upon localized regions of tissue. Steps in this direction have been taken by Murphy and Bennett (10). Certainly, the levels encountered in this study were not sufficient to elicit complaints from the subjects. They do not represent long-term application and hence would not include cumulative effects.

In dynamic alignment, the criterion of judgment of proper alignment is a good gait pattern. This is a matter of sensation of balance, which may or may not be affected by pressure-sensation feedback. The probable use of interface pressure as a predictor of sound gait seems remote.

CONCLUSION

This study leads one to conclude that the pressures at the critical points examined for this set of subjects showed considerable variation and exceeded the estimated nominal uniform normal pressure. For both static stance-support modes and maximum dynamic pressures of walking, the values for the distal anterior tibial region tended to be the largest and most sensitive to alignment alterations. Pressure variation appeared to be more sensitive to angular alignment than to linear alignment changes. Judging from patient and stump reactions to the tests, the pressure levels appear to be well within tolerance levels.

The static pressure during support by the prosthesis only with contracted stump muscles approaches that of maximum dynamic pressure and might be used as a clinical measurement to provide protection against excessive dynamic pressures.

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