END-BEARING CHARACTERISTICS
OF PATELLAR-TENDON-BEARING PROSTHESSES —
A PRELIMINARY REPORT

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ABSTRACT

The present work was aimed at investigating the degree of participation of the stump end in the weightbearing process in PTB prostheses. Despite the belief that the stump end is not capable of withstanding high loads, it was revealed that most patients can bear from 15 percent up to 45 percent and more of their weight on the stump end. The limited number of patients studied diminishes the statistical significance, but the results of this work do suggest that the weightbear-
ing mechanism of a PTB prosthesis can be further improved by transferring more load through the stump end. It is suggested that a non-destructive technique for assessing the end-bearing efficiency of the prosthesis be developed. Also, the criteria for design of the socket end in order to produce the desired load-bearing features needs further investigation.

**INTRODUCTION**

Fabrication procedures of below-knee (BK) prostheses have been considerably improved during the last two decades. Today, unless other health complications are involved, BK amputations hardly impose functional restrictions. However, since the proper fitting of prostheses is greatly dependent on the skills of the fitter, some failures may occur.

The introduction of the patellar-tendon-bearing (PTB) concept by Radcliffe and Foort during the late fifties represents the greatest contribution to the advancement of the below-knee prosthesis. The PTB prosthesis is based on the concept of transferring load via specialized regions, as opposed to the concept of homogenous load distribution on the stump/socket interface. The weightbearing forces are conveyed through specialized regions on the stump such as the patellar tendon, the tapering regions of the tibial condyles, and the relatively horizontal portion of the entire stump. The load-sharing between the various regions varies alternately in accordance with the angular position of the shank. Stabilizing and ambulating moments are transferred through the medial and lateral condyles of the tibia and through the distal one-third of the lateral side of the leg. The biomechanical adequacy of the PTB concept was later investigated and approved by Sonck et al. (1970), and Pearson et al. (1973).

Despite the success of the PTB concept, there are occasional complaints of painful reactions and skin irritation or mild ulceration in the main loadbearing regions. To overcome the problem, several attempts were made to further optimize the load distribution (Stewart, 1970).

The stump end was traditionally considered to be very sensitive to loadbearing and therefore was expected to bear a negligible fraction of the weight, although mechanically the stump end can bear weight most efficiently, since it presents a surface that is almost perpendicular to the force vector. The PTB fabrication instructions indicate that total contact must exist between the socket and the stump.

A few attempts have been made to develop methods of assuring correct execution of the PTB concept. These were related to observa-
tion of the total contact as well as load distribution. The use of talcum powder in the socket to detect interface clearances is quite popular in prosthetic shops. Microcapsules of staining substance are also occasionally used to provide some indication of pressure distribution. (The microcapsules burst under increased pressure and stain the socket. This technique is rather inconvenient.) A ball of clay is used to detect possible clearance between the stump and the socket. Also radiography (Taft, 1969) and thermography (Faulkner, 1973) have been employed for the same purpose, but these can be applied only in limb-fitting centers which have access to those facilities. Transparent sockets as described by Grille (1969) can be rather useful, particularly when no soft lining is used between the stump and the socket. All these techniques give only a qualitative impression rather than a quantitative measure of the adequacy of the fitting.

Not much information is available as to the amount of load which can be transferred via the stump end. A successful attempt to increase the weightbearing on the stump end was made by Wilson, Lyquist and Radcliffe (Wilson, 1968) by introducing the air-cushion socket. The elasticity of the inner socket and the air cushion provided a moderate and evenly distributed pressure on the stump end. Lately, the resilient foamed silicone rubber pad has been introduced for filling up spaces between the stump end and the socket; this also allows a certain amount of control for end-bearing. The two techniques are meant to ensure total contact and a certain amount of end-bearing. However, they are not based on the actual amount of load which the stump end can tolerate, and as we have noted, not much information is available as to the limits of load which can be transferred by the stump end. That type of information will obviously depend on the structure of the stump, the important factors being amount of soft tissue at the distal end, the cross-section area of the bone, and the general state of health of the stump tissue and its sensitivity. The purpose of the present work is to evaluate the loadbearing capability of the stump end (on a limited number of patients) and to propose criteria for achieving the correct load transfer.

**MATERIAL AND METHODS**

**Subjects**

Six patients (five male and one female) were investigated. Age range was 27–67 yrs. All subjects were in a state of general good health with well-stabilized stumps. The stump dimensions of all patients were within the standard range of 15–20 cm long and no excess
soft tissue was observed at the distal end, except for patient No. 3 who had a rather "fleshy" stump.

Three patients were less than 1 yr postamputation and the other three were 4 yrs and more postamputation. The following parameters were recorded for each subject: weight, sex, date of amputation, and cause for amputation (Table 1).

<table>
<thead>
<tr>
<th>No.</th>
<th>Age yrs</th>
<th>Sex</th>
<th>Weight (kg)</th>
<th>Cause of amputation</th>
<th>Side</th>
<th>Time elapsed since amputation (yrs)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>65</td>
<td>M</td>
<td>65</td>
<td>A.S.V.D.</td>
<td>Right</td>
<td>5/12</td>
</tr>
<tr>
<td>2</td>
<td>64</td>
<td>M</td>
<td>57.5</td>
<td>A.S.V.D.</td>
<td>Right</td>
<td>4/12</td>
</tr>
<tr>
<td>3</td>
<td>35</td>
<td>F</td>
<td>77.5</td>
<td>Tumor</td>
<td>Right</td>
<td>9/12</td>
</tr>
<tr>
<td>4</td>
<td>28</td>
<td>M</td>
<td>80</td>
<td>Trauma</td>
<td>Right</td>
<td>5</td>
</tr>
<tr>
<td>5</td>
<td>32</td>
<td>M</td>
<td>67.5</td>
<td>Trauma</td>
<td>Right</td>
<td>10</td>
</tr>
<tr>
<td>6</td>
<td>50</td>
<td>M</td>
<td>56</td>
<td>Trauma</td>
<td>Right</td>
<td>3</td>
</tr>
</tbody>
</table>

**Experimental Socket**

Experimental sockets were manufactured according to the standard PTB instructions as prescribed by Radcliffe and Foort (1961). The actual fitting procedures are influenced by the instructional course delivered by New York University (the Postgraduate Medical School). All sockets were conventional PTB sockets fitted to one full-thickness stump sock with strap suspension and a Pelite soft lining. The upper third of the socket was fixed by four flat metal straps to a wooden plate. The latter was connected to the proximal end of an Otto Bock modular shank (Fig. 1). Two tubular synthetic lugs were attached to the distal two-thirds of the socket, parallel with the longitudinal axis of the socket. A transverse saw cut was then made through the distal end of the socket; the cut also passed through the middle of the lugs.

Metal guide rods were mounted in the axially-oriented holes of the lugs and fixed to the top half. The rods and lugs thus connected the two parts of the socket, allowing free movement of the distal end of the socket along the prosthesis axis. When the patient wore the prosthesis, the distal end of the socket became compressed against a load cell (installed on top of the wooden plate) that measured the load transmitted by the distal end of the stump.
The end-bearing measurements were made by using a Kyowa LM-50-KA load cell (20 mm dia, 9.5 mm thick and 20 g weight). The axis of measurement of the load cell coincided with the longitudinal axis of the socket. The location of the load cell enabled weightbearing measurements of the distal end of the socket, when the socket was pressed by the distal end of the stump. The load cell was calibrated prior to the experiments: good linearity with low drift was achieved.

The Measuring System

End-Bearing Load Cell.

FIGURE 1. — (A) The experimental prosthesis, lateral view.
Load variation comparison between the distal end and the entire stump was effected with two force-plate dynamometers. Each subject was asked to walk at a comfortable pace along the walk path, which included two built-in Kistler z-3482 piezoelectric force platforms. The vertical force channel of the platform provided a reasonable approximation of the total axial force transmitted from the ground to the prosthesis. This approximation is based on the fact that the angular inclination of the shank in relation to the vertical, during the main
weightbearing phase (excluding the double-support phase), ranges from $+15$ deg to $-15$ deg. The cosine of this angle is close to 1 and sine is 0.25. However, the sine affects the contribution of the horizontal shear (A–P. component) which is several-fold smaller than the vertical component. The total error resulting from this assumption, therefore, will be less than 10 percent of the actual axial force in the extreme positions of 15 deg. inclination. The measuring system (load cell and force platforms) was connected via a system of amplifiers to a multichannel ultraviolet recorder. Simultaneous characteristics of the forces from the two measuring systems were recorded for further comparison and processing. The analog force characteristics were later digitized with a Graph-Pen digitizer and integrated to obtain the total impulse values.

**Test Procedure**

A narrow spacing ring was cut from the distal end of the main part of the socket to allow augmentation of load on the stump end by pressing the distal end of the socket proximally against the bottom of the stump. If further removal of material was required to increase the loading of the stump end, this was done prior to consecutive tests. The final shortening of the socket length following this procedure could amount to 2 cm. However the actual clearance between the two parts of the socket never exceeded several millimeters and therefore no tissue bulging or pinching occurred.

The load cell and force plates output readings were zeroed. Each patient was then asked to perform the following procedures with the prosthetic device in place:

1. Stand on the force plate with only the prosthetic foot bearing the full body weight.
2. Walk at a normal cadence along the walk path with the aim of treading on the force plates.

The first test was carried out with the socket at its original dimensions, e.g., the bottom of the socket was placed at the correct distance as obtained from the original cast. Following every successful measurement, another metal disk 2-mm thick was placed between the distal end of the socket and the load cell to increase the end-bearing force. The repeat measurements were recorded. Spacers were piled one on top of the other and measurements were repeated until the patient complained of pain at the distal end of the stump.

**RESULTS AND DISCUSSION**

As described above, the forces on the stump end and those acting
on the foot were simultaneously monitored. Our main focus was upon the axial component of force applied to the foot by the ground during the single-leg support. It was therefore assumed that the vertical ground reaction component represents the axial force on the leg during this phase with a very good approximation, as discussed in the previous section.

The first test performed with each patient was aimed at examining the degree of total contact obtained during the new socket fabrication procedure. This was done, as described above, by placing the bottom section of the socket at its exact axial distance as manufactured. The results of the end-bearing measurements revealed that in four patients there was no end bearing at all. The other two patients had relatively low end bearing of the order of 5 kg. These two results are more or less in agreement with the total-contact (but not end-bearing) concept. Following the first test the bottom of the socket was elevated toward the main part by inserting the 2-mm spacing washers between the load cell and the socket end. This was done without removal of the prosthesis as long as the clearance between the parts of the socket allowed for such adjustment. If the gap was too narrow, the prosthesis was dismantled for additional removal of material. (This procedure may have biased the experimentation to a limited extent due to swelling of the stump, but no real swelling was observed or noted by the team or the amputee.)

This procedure gradually increased the load on the stump end. After every such respacing, the forces were recorded. The aim of this procedure was to detect the ultimate compression possible without causing pain and inconvenience. During this procedure two important readings were evaluated: the last measurement obtained without any pain, and the first measurement obtained with pain. Since our socket-spacing procedure was intermittent rather than continuous, the pain threshold should lie between those two values.

Typical results, as obtained from two patients, are given in Figure 2. The interrupted line describes the force acting on the end of the stump as measured by the load cell. The solid line describes the vertical component of the ground reaction force. Figures 2a and 2b relate to the patient No. 3: maximum force without pain, and the next measurement with corresponding pain. Figures 2c and 2d relate to patient No. 4, who can withstand a very high load on the stump end. The differences between the interrupted line values of Figures 2a and 2b, and between 2c and 2d, give the pain threshold range for patients No. 3 and No. 4, respectively. The rather extended force characteristic obtained from the results of end-bearing on patient No. 4 represents a residual force action on the stump end even when the
prosthesis is no longer bearing weight. This is a kind of “prestressing” applied to the stump by the shortened socket. It is interesting to note, in the results obtained on patient No. 4 as seen in figure 2c and 2d, that at this high level of compression of the stump the soft tissue behaves almost as non-compressible fluid. A minor displacement of 2 mm results in a 12-kg increase of the end-bearing force.

Due to the nature of the experiment in which the displacement is controlled intermittently, the actual pain threshold can be defined only within these broad limits. Within the limits of accuracy of these experiments in relation to the sensation of pain by the patients, it was evident that pain occurred rather sharply at a certain level of end-bearing force. This was contradictory to the assumption that pain will develop gradually with the increase of weightbearing, from slight inconvenience up to a sharp and unbearable pain. However, the use of these prostheses was confined to the laboratory; it is anticipated
that pain may develop at lower levels of compression during prolonged use. The results of the pain threshold range for all patients, in the standing position and during gait, are given in Table 2.

<table>
<thead>
<tr>
<th>Patient No.</th>
<th>Maximum end-bearing in standing</th>
<th>Maximum end-bearing in gait</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Without pain</td>
<td>With pain</td>
</tr>
<tr>
<td></td>
<td>% of vertical</td>
<td>% of vertical</td>
</tr>
<tr>
<td></td>
<td>Value kg force</td>
<td>Value kg force</td>
</tr>
<tr>
<td>1</td>
<td>13.5 21</td>
<td>22.5 35</td>
</tr>
<tr>
<td>2</td>
<td>18 31</td>
<td>22.5 39</td>
</tr>
<tr>
<td>3</td>
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<tr>
<td>4</td>
<td>34.5 43</td>
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<tr>
<td>5</td>
<td>19 28</td>
<td>22 33</td>
</tr>
<tr>
<td>6</td>
<td>26.5 48</td>
<td>34.5 63</td>
</tr>
</tbody>
</table>

The range is defined by the maximum measured loadbearing without pain, and the next value which produces pain. The forces are given in kg and in percent of the total weightbearing on the leg.

It can be seen from Table 2 that patients No. 4 and No. 6 could withstand stump-end loads which exceed 50 percent of their total weightbearing. This is even higher in terms of percentage of body weight.

It was assumed that another significant parameter which could help in defining the end-bearing characteristics of the stump was the rate at which the force developed towards its climax. This was measured in terms of the time it takes for the force to reach 90 percent of the maximum value. The 90 percent value was chosen to avoid searching for the climax in the slight fluctuations of the force characteristics at the steady-state range.

It can be seen in Table 3 that, in most cases, the force reaches a steady-state value in the time range of 13–35 percent of the stance phase. Those figures are 8–21 percent of the total walk cycle. This means that in two of the cases (No. 4 and No. 5) the force reaches its peak value within the double-support phase. In the other cases the maximum force development is delayed beyond this phase. This is associated with a certain piston action caused by soft-tissue delayed compression. No correlation was found between the last parameter and the body build or tightness of socket fit. It therefore seems that the relatively rapid force buildup in cases No. 4 and No. 5 is associated with the impulsiveness of gait.
TABLE 3. — Time Percent of the Stance Phase Taken to Reach 90 Percent
Maximal Force Value.

<table>
<thead>
<tr>
<th>Patient No.</th>
<th>Maximum force kg</th>
<th>Time percent to 90 percent force</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>13</td>
<td>22</td>
</tr>
<tr>
<td>2</td>
<td>16.5</td>
<td>30</td>
</tr>
<tr>
<td>3</td>
<td>8</td>
<td>35</td>
</tr>
<tr>
<td>4</td>
<td>39.5</td>
<td>14</td>
</tr>
<tr>
<td>5</td>
<td>22.5</td>
<td>13</td>
</tr>
<tr>
<td>6</td>
<td>32</td>
<td>29</td>
</tr>
</tbody>
</table>

Figure 2 is typical of 12 such characteristics obtained on the 6 patients. The force is plotted against time, and the force-time integral gives the total value of the impulse. This is represented by the area enclosed between the force characteristics and the time axis. All the force characteristics were normalized and integrated. The normalization was carried out by dividing the force coordinate by body weight, and time coordinate by the total stance time. As a result, a nondimensional figure for the impulse is obtained. In the (static) standing position this figure will be unity for a specific interval, regardless of the length of time. The ratios of the impulses transmitted through the stump end to the total impulses transmitted to the body, for each individual, are given in Table 4. The impulse is an important value since it provides an integrated measure of the force transmission to the stump. The impulse ratio values differ from the maximum force values given in Table 2 by definition. The force values may represent a local phenomenon while the impulse represents the overall effect. The normalized impulse can be larger than unity or smaller than unity. Large values of impulse will indicate efficient use of the prosthetic leg and “impulsive” gait, while low values will represent a rather inefficient use of the prosthesis. In other words, low values of impulse will be representative of a familiar tendency of the amputee to “rock over” the prosthesis while using mainly his good leg for support. This phenomenon is independent of the actual value of the peak force reached, which might have lasted for a very short period of time.

It can be seen clearly from Table 4 that patients No. 1, No. 2, and No. 3 belong to one category while patients No. 4, No. 5, and No. 6 belong to another category. No significant correlation was found with body build or cause of amputation. The only common factor among patients No. 4, No. 5 and No. 6 was the time elapsed since the amputation; the first three patients were less than a year postamputation, while the other three had undergone amputation at least 4 yrs ago.
Another finding, which seems to be of a more general nature and not specific to the present study, was the stance-to-overall-walk-cycle time ratio. From all measurements performed (four to six tests per patient), it was evident that the stance time was reduced with the increase in stump end-force. This was initially thought to be associated with the tendency to reduce the weightbearing period on the prosthesis. However, further examination of the results revealed that the ratio of stance time to the time of the full cycle remained constant (see Table 5). In other words, the patients increased their pace of walking rather than change the times of support of the two legs. This, although irrelevant to the present study, indicated that increased pressure or pain does not immediately alter the task-sharing between the two legs.

**SUMMARY AND CONCLUSION**

The present work was aimed at investigating the degree of participation of the stump end in the weightbearing process in PTB prostheses. The force/time characteristics of the stump end and of the leg as a whole were recorded.
It was found that as a result of the socket fabrication procedure employed, only two sockets were total-contact sockets. This is due to lack of an appropriate nondestructive test technique to control the end-bearing characteristics during fabrication.

Despite the belief that the stump end is not capable of withstanding high loads, it was revealed that most patients can bear from 15 percent up to 45 percent and more of their weight on the stump end. Naturally, the limited number of patients studied diminishes the statistical significance of these results. It seems that old (in terms of postamputation time) amputees, especially with traumatic cause, are less sensitive and can withstand 30 percent and more weight on their stump end.

The results of this work suggest that the weightbearing mechanism of a PTB prosthesis can be further improved by transferring more load through the stump end. This will reduce the number of complaints about painful pressures on the patellar tendon and the other loadbearing regions.

It is suggested that a nondestructive technique for assessing the end-bearing efficiency of the prosthesis be developed.

The criteria for design of the socket end in order to produce the desired loadbearing features needs further investigation. Our findings, based on axial shifts of the bottom of the socket towards the stump, and their correlation with the loads produced, provide only a fraction of the required information.

The results presented in this preliminary study are representative of only a short-term use of the prosthesis within the laboratory. The effect of prolonged use of the prosthesis, under varying levels of end-bearing, upon the development of pain is the subject of our following investigation.

The present use of a split socket is merely for experimental purposes. The ultimate goal of the continuing study could be to reach criteria for the prescription of the individual socket length, based most likely on the stump configuration, position of the tibial end, amount of distal soft-tissue padding, and sensitivity to end pressures on the part of the amputee.

**ACKNOWLEDGMENTS**

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REFERENCES


