The Seattle Prosthetic Foot—A Design for Active Sports: Preliminary Studies

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INTRODUCTION

The concern of many amputees is focused not only on the basic requirements of daily living, but also upon the quality of life as expressed in sports and recreation. A survey of 134 lower limb amputees indicated that 61% participated in some type of recreation, yet only a few of these individuals wore specifically designed recreational prostheses.[6,7,8,9] Many expressed a lack of interest and/or lack of knowledge on the part of prosthetists and other members of the amputee team regarding the need and desire of amputees to participate in physically active recreation.

The ability to exercise naturally affects quality of life. The general concern for improved physical fitness has, during the last few years, prompted national measures to upgrade the level of citizen exercise. For the physically handicapped, this need is even more important. In the case of the amputee, a considerable part of physical performance is related to the prosthesis. This is especially so with younger, vigorous persons. (Fig. 1).

The major specific problem encountered in sports was running, followed by walking long distances. Improved prosthetic design was indicated as the primary requirement to expand physical capability.

Fig. 1. Prosthetic design can affect the performance of a physically active amputee.

ANALYSIS OF RUNNING FOR BELOW KNEE AMPUTEES

The ability to run opens up a wide vista of sports for lower limb amputees. There is a common consensus among amputees that they are unable to run. Therefore few amputees participate in vigorous physical activities which may require this level of participation. There was little objective information on amputee running, thus, three years ago we undertood a research project directed to identifying the kinematics and kinetics of amputee running performance.[2,3,4] A collaborative study with the authors and Dr. Doris Miller and her associates at the University of Washington, Department of Kinesiology was designed to determine how successfully unilateral below knee amputees could run. Data was accumulated in the preliminary study of ten subjects.

Front, back, and side views of physically active candidates, running at self-selected constant speeds, were filmed at 100 frames per second with a LOCAM camera positioned with the optical axis perpendicular to the plane of motion. The measurement of ground reaction forces acting on the prosthesis during running was recorded using a Kistler forceplate. The subjects wore gym clothing and shoes routinely used for sports participation. The running gait was filmed and data stored on magnetic tape, with computer reduction and analysis[10] (Figs. 2 and 3).

A number of noteworthy observations were made. For example, a few of the amputees were not aware that they could run; however, after several sessions they found that they were more mobile than they had been before the training. In fact, one unilateral below knee amputee was able to run forty yards in five seconds after coaching.

The most common undesirable characteristics exhibited by the ten subjects were: (1) maintenance of an excessively straight knee (locked knee) on the prosthetic side during heel contact, which reduced the shock absorption function of the residual limb and placed unnatural stress on the knee, hip and vertebral column and, (2) restricted range of motion of the intact limb at the knee and hip during swing phase. Recovery of the limb with so little knee flexion could only be accomplished by additional contraction of the quadriceps muscles resulting in unnecessary fatigue.

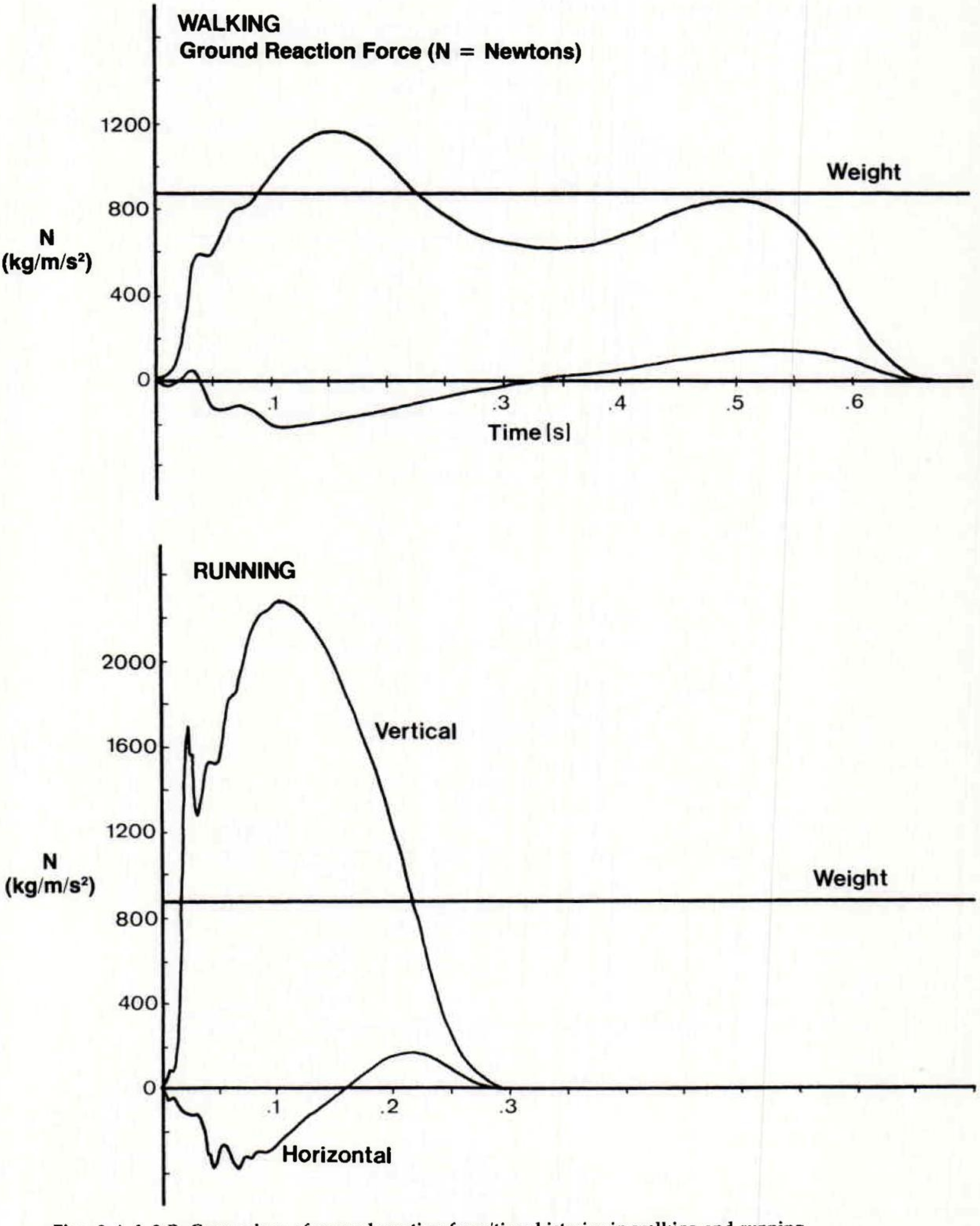
As was expected, there were marked differences between the intact and prosthetic feet, particularly in the ranges of plantar flexion and dorsiflexion. Normal foot function during running also requires a pronation/supination factor which allows the foot to roll inward on the lateral border after contacting the ground. This component is not adequately built into most prosthetic feet.[11]

Design, alignment, suspension, and materials also contribute to successful below knee "running" prostheses. Preliminary studies show that running prostheses should not be set in 5 to 10 degrees of flexion as compared to a conventional prosthesis, but should be plantar flexed so that the runner's weight can be centered over the ball of the foot during pushoff.

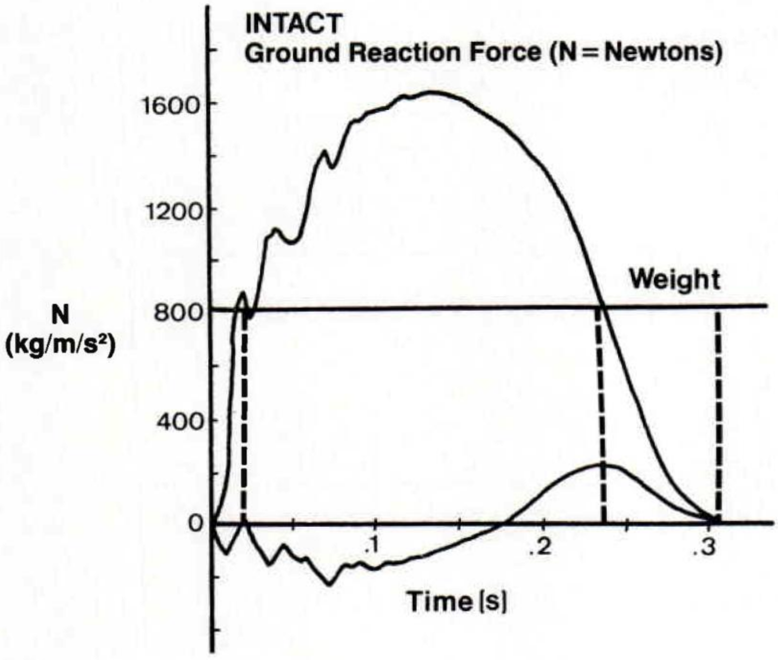
Several amputees preferred rubber latex sleeves or similar suspension straps to minimize pistoning while running.

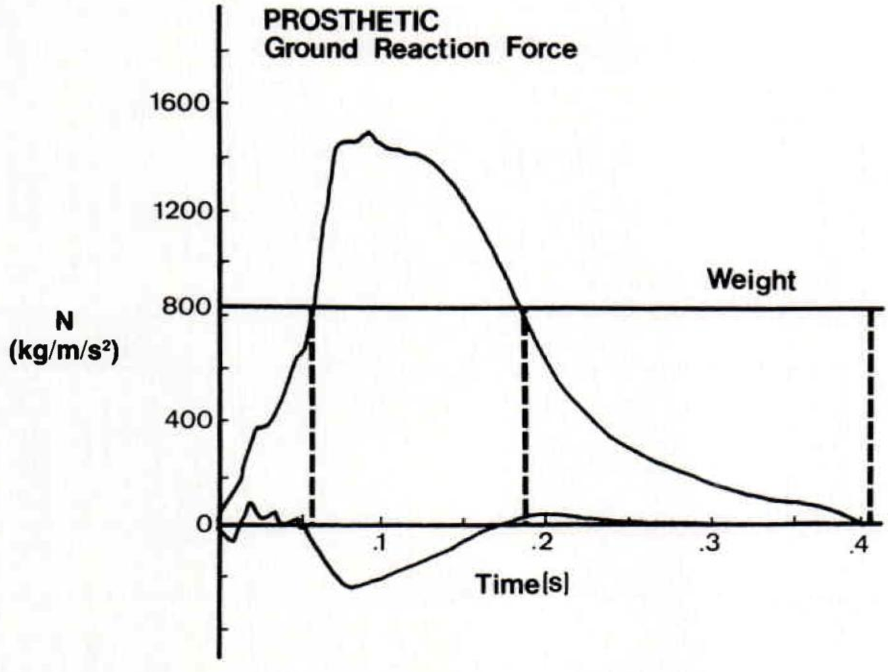
Since the conventional prosthesis is intended for walking, the vertical ground reaction (or impact force) is rarely much greater than body weight. In running, however, this force reaches two to three times body weight. The net effect of this mismatch between design specifications and utilization is two-fold: a shortening of the life of the prosthesis and the development of a gait which is potentially damaging to lower limb joints.

In our analysis of the prosthetic requirements to improve performance, the need for an energy storing ankle/foot design became evident. Using the data from the kinesiology studies, we have designed a foot incorporating a leaf-spring mechanism to aid in pushoff, imitating the activity of the gastrocnemius and soleus muscles. The spring stores and releases the energy of gravitational compression for the purpose of enhancing running performance (Fig 4).[1,5]



Figs. 2-A & 2-B. Comparison of ground reaction force/time histories in walking and running.





Figs. 3-A & 3-B. Ground reaction component force/time histories during intact (above) and prosthetic stance (for one subject). Solid bars designate double support while the vertical arrows indicate the beginning of knee flexion prior to take-off.

Fig. 4. The leaf spring in the foot stores energy by compressing as weight is placed upon it.

PROSTHETIC DESIGN

To be practical, the foot needs to be simple in design, lightweight and durable. Since action depends upon the spring assembly, certain constraints with regard to materials and their performance need to be made. Force requirements and the amount of deflection during force application become critical. It was through running analysis that the optimum deflection angle of 22 degrees, or 1¾ inches, was determined. Given the load and the amount of deflection, the number of leaves in the spring could be calculated. After comparing the weight of each material and the performance requirements together with the need to minimize weight, fiberglass was selected.

It became evident through testing that one spring alone could not provide enough downward force during pushoff. The design initially utilized contained three springs with a rubber deflection bumper. Problems emerged, however, specifically at the point of attachment where repetitive testing caused spring breakage. After a series of additional design modifications, including extensive bench and field testing, the present design emerged (Fig. 5A & B).

This design proved effective with one exception; if forcibly plantarflexed, the foot could delaminate. Delamination oc-

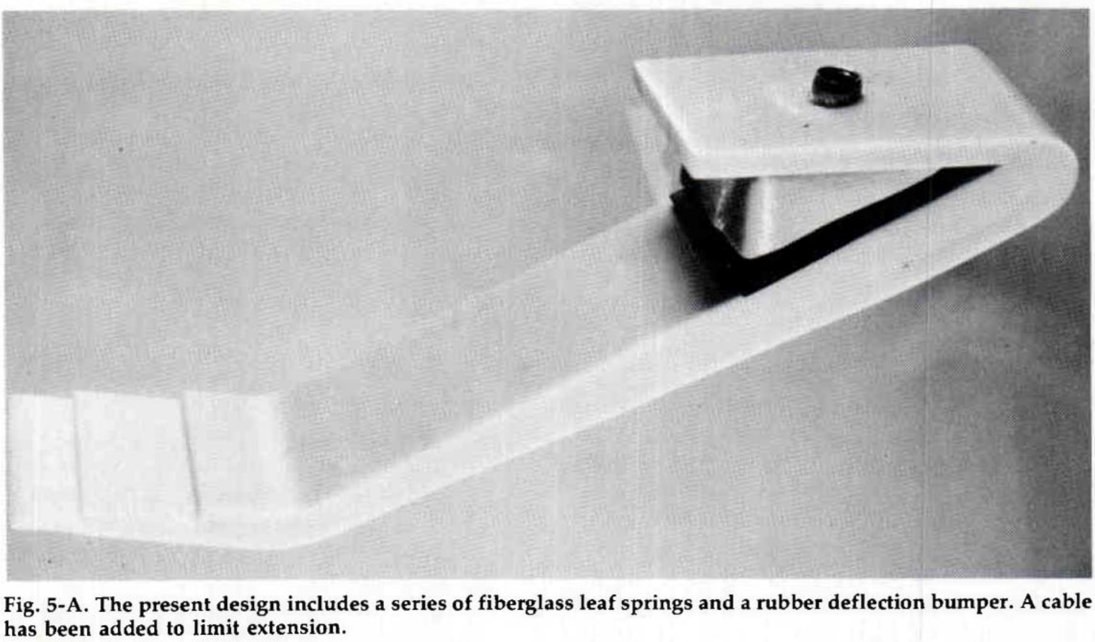


Fig. 5-A. The present design includes a series of fiberglass leaf springs and a rubber deflection bumper. A cable has been added to limit extension.

curred while one of our subjects was skiing downhill. He leaned back in an exaggerated position causing the lever arm of the ski to forceably extend the foot. The problem was solved with the addition of an extension limitation cable added just anterior to the deflection bumper, allowing the foot to compress but not extend.

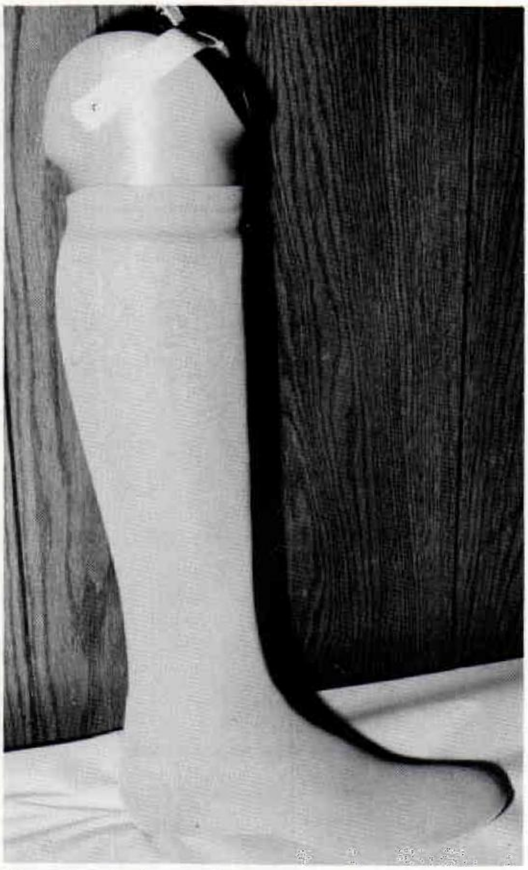
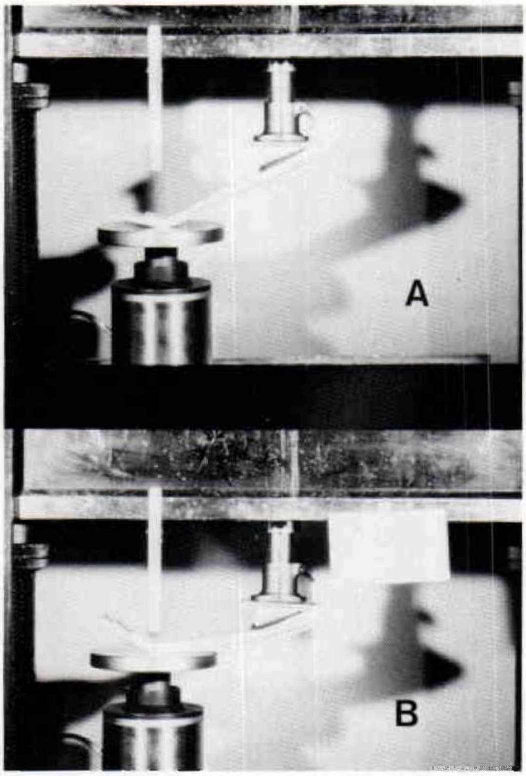


Fig. 5-B. The foot is finished to appear like a normal SACH foot.

PERFORMANCE

Performance of the foot is being evaluated first through laboratory testing. The strength in the spring assembly is tested on a machine which records the amount of deflection measured against the force supplied to the point of breakage (Fig. 6A & B). The second performance evaluation is running analysis through the Kistler forceplate studies. Since most of the amputees



Figs. 4-A & 4-B. Testing force related to deflection in order to determine performance characteristics and the failure point.

have endoskeletal systems allowing alignment and foot changes, the Kistler forceplate provides a good means of comparison not only against the normal but against other prosthetic components. The third performance evaluation is through criticism of the wearer (patient response).

The patients' response has been varied with the following comments being characteristic:

1. On the average, the patient got used to the foot in approximately one week.
2. Initially, some of the patients had difficulty with slow walking. The foot tended to throw the leg forward, but through continued use, this problem seemed to alleviate itself.
3. Running was easier with increased stride length and pushoff. This is accomplished through the foot's ability to store and release energy.

4. It made ascending ramps and stairs easier because it offered the patient more pushoff.

5. Generally, the foot increased one's activity level to include those activities such as running, jumping, etc., with increased ease and comfort.

6. Some patients felt a psychological attachment to the foot and were unwilling to give up its use because of the improved function and performance it provided them.

While the foot's performance has allowed amputees to participate in a broad and increasing vista of activities through the storing and releasing of gravitational energy, research is continuing to improve overall function. There are also plans to modify the energy storing characteristic so that the design and performance characteristics can be included for routine walking activities.

CONCLUSION

The Seattle prosthetic foot design presented is a combination of materials and engineering knowledge. It has been constructed to dynamically store and release energy through controlled spring motion. This preliminary report outlines our progress to date.

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