LOCOMOTION ASSISTANCE THROUGH CANE IMPULSE

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ABSTRACT

The extent of locomotion assistance gained through cane usage by those with hip disorders is assayed. Employing as a standard the propulsive impulse delivered by each lower limb of a healthy young male, the cane is shown to supply about one-fifth the equivalent impulse, aside from other possible benefits. Full test values are given for nine handicapped subjects.

INTRODUCTION

Use of a cane is known to act in one or more planes to increase stability during locomotion, supply sensory information, reduce the

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Bennett et al: Locomotion Assistance Through Cane Impulse loading of musculoskeletal structures, and contribute to the acceleration and braking requirements of bipedal locomotion. This study quantitates one aspect of cane use by computing the amplitude of the accelerative and braking cane impulses during locomotion for nine subjects with hip pain.

Materials and Methods

Nine men, all of whom had no disability other than severe pain in one hip resulting from degenerative joint disease and/or avascular necrosis, participated in the study. Their ages, heights and weights are given in Table 1. All of the men were experienced cane users, and each used his cane in the hand opposite the side of the pain. When walking without a cane, all subjects exhibited the pain-avoidance maneuvers characteristic of the antalgic limp associated with hip pain (1).

<table>
<thead>
<tr>
<th>Subject</th>
<th>Age (In)</th>
<th>Height (In)</th>
<th>Weight (Lbf)</th>
<th>Ave. Vel. (Ft/Sec)</th>
<th>Ave. Axial Force (Lbf-Sec)</th>
<th>Cane Contact Time (Sec)</th>
<th>Cane Brake Impulse (Lbf-Sec)</th>
<th>Cane Accel. Impulse (Lbf-Sec)</th>
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<tbody>
<tr>
<td>1</td>
<td>44</td>
<td>70.5</td>
<td>176</td>
<td>2.60</td>
<td>14.2</td>
<td>0.85</td>
<td>0.39</td>
<td>0.59</td>
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<tr>
<td>2</td>
<td>69</td>
<td>67</td>
<td>185</td>
<td>2.70</td>
<td>15.5</td>
<td>0.58</td>
<td>0.11</td>
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<td>3</td>
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<td>66</td>
<td>223</td>
<td>1.53</td>
<td>10.6</td>
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<td>1.05</td>
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</tr>
<tr>
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<td>145</td>
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<td>12.2</td>
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<td>70</td>
<td>150</td>
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<td>22.0</td>
<td>0.70</td>
<td>0.58</td>
<td>0.79</td>
</tr>
</tbody>
</table>

The cane force data reported in this study were obtained from an instrumented cane developed by Seireg, et al. (2), which measured the axial force as a function of time. The displacement patterns of the body and the cane shaft during walking were obtained with a previously described method of interrupted-light photography at the Wood Veterans Administration Medical Center, Wood, Wisconsin (3). For each subject, the following measurements were made: the amplitude of the axial force applied to the cane, the orientation of the cane in the sagittal plane throughout the period of cane contact, and the average velocity of the subject while walking with the cane. A typical example of a pattern of cane force versus time is shown in Figure 1 together with a plot of the fore-aft orientation angle of the cane shaft with the vertical during the same time period.
Computation of Braking and Accelerative Impulses

The average momentum of each subject during walking was calculated as \( \text{Mom} = \frac{wv}{g} \) where weight \( w \) in lbf divided by the acceleration of gravity \( g \) is the mass, and \( v \) is the subject's mean forward velocity. For example, if the value of 4.5 ft/sec is used as representative of the average forward velocity of normal young men (4), along with a weight estimate of 150 lbf, it may be shown that the corresponding momentum value is approximately 21 lbf-sec. Actually there are variations above and below the average momentum in each stance phase of walking because velocity varies rhythmically.

For the purpose of this study, the variation in momentum during the stance phase of a single limb is considered and this is diagramatically shown in Figure 2. At the instant of heel contact, a braking process is initiated in which the basic forward momentum is considerably reduced. The braking process continues until the leg is...
substantially vertical, with the foot flat on the floor (5), and forward momentum is at a minimum. The ensuing push-off process increases momentum until toe-off when the entire initial momentum is restored. Thus, the basic momentum level is continuously altered as a function of time in approximately a sinusoidal pathway about the average value.

To compute the amplitude of the momentum wave resulting from the action transmitted through a single lower limb, the classic impulse-momentum relationship was used. The change in momentum (i.e., the net impulse) is given by the integral of the cane fore-and-aft shear force $F$ with respect to time: $\Delta \text{Mom} = \int F \, dt$.

Figure 3 shows a force plate record of the fore-and-aft shear forces of a normal young man during the stance phase of locomotion (6), walking without a cane. These forces are used in calculating changes
in forward momentum. The initial or negative area phase of the record corresponds to the braking tendency following heel contact, and the positive area portion corresponds to the momentum increase developed in push-off.

Analysis of the ground reactions of three normal subjects (6) by conventional graphical strip-integration procedures yielded an average value of 4.3 lbf-sec for the braking portion of the stance phase and 4.8 lbf-sec for the accelerating portion. As a practical measure a rough average value, 4.5 lbf-sec, will be used to represent the single lower limb impulse developed by normal young men in the course of the braking or accelerating phase of gait. The value 4.5 lbf-sec shall be called a "limbworth" of impulse and shall be used as a basis for comparing cane output.

Cane impulse was computed by $\int F \sin \theta \, dt$ through strip integration of the force-time curve. $F$ is the cane resultant (axial) load, $t$ is the time of ground contact and $\theta$ is the fore-aft cane angle from the vertical. Positive and negative values of $\theta$ indicate acceleration (positive) or brake (negative) phases of load transfer (Fig. 1). Correspond-
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ingly, the decrease in momentum from heel contact to footflat (Fig. 2) is caused by the brake phase momentum represented by the area of the negative loop in Figure 3.

RESULTS

Table 1 presents individual data for average velocity, average momentum, cane contact time, and cane braking and accelerating impulses for the nine men with hip pain. The average cane braking impulse developed by the subjects was 0.66 lbf-sec or 14 percent of a limbsworth of impulse. For eight of the nine subjects the accelerative impulse was greater than the braking impulse. The accelerative impulses averaged 1.41 lbf-sec or 31 percent of a limbsworth of impulse. Neither the braking nor accelerative impulses correlated significantly with body weight. In Figure 4 each subject’s braking and accelerating impulses are depicted in order of decreasing acceleration impulses, and compared to the amplitude of a single limbsworth of impulse.

DISCUSSION

The results reported in this study must be viewed in light of the limitations of the technique. This study measured the direct contribution to forward momentum produced by the cane and only by the cane. The alterations of the normal momentum interchange in the body segments of the subjects because they were using the canes are not considered. Any comprehensive study of momentum would require full data from two force plates and measurements of instantaneous velocity of the subject, as well as data on the cane orientation and forces applied to the cane.

In order to achieve a reasonably constant-velocity gait, normal subjects must generate or transmit approximately equal amounts of accelerating and braking impulses. The cane’s function is unlike that of a lower limb during gait, in that the cane user has a free choice as regards the orientation of the cane and the magnitude of the impulse imparted to the cane. In principle, the cane can be used to generate an upward acceleration only, an accelerating impulse only, or a braking impulse only, (or any combination of such impulses) and the cane user may still maintain a constant-velocity gait. In practice, since the cane subserves the needs and deficits of the user, he trades-off many variables such as pain, high energy cost, and instability to arrive at a combination he deems optimal. Without seeking to evaluate the wisdom of each such solution, this paper addresses itself to the propulsive and restraint functions of cane use.

The cane users in this study chose to develop an average of 14 percent of a limbsworth of braking impulse and 31 percent of a limbsworth of accelerative impulse.
limbsworth of accelerating impulse. Thus, the accelerating aspect of cane use in these patients with hip pain was employed more extensively than the brake aspect by an amplitude of approximately two to one. (The "limbsworth" unit is based on the action of the lower limbs of healthy young men in accelerating and braking during gait; most of the subjects in this study were middle-aged or elderly.) It is apparent that the use of a cane for propulsion requires a great deal of muscular effort from the upper limb of the user. Thus, the use of a cane for propulsion is seen as highly significant to the user; were it not so, he would not make the effort.

Cane use subserves more than propulsion and restraining functions (7). Since the cane angles of the subjects in this study never exceeded 20 deg from the vertical, it is apparent that a large part of their
applied force was also effective in decreasing the vertical load on the painful hip. As shown in Figure 5, the cane is acting at a long lever arm so that even a modest force can produce a large moment at the hip joint center. This would reduce the amount of force required from hip abductor musculature, which normally contracts during single-limb support to control the descent of the pelvis on the unsupported side (1, 7-11). The load relief derived in this fashion can make walking tolerable to many who would otherwise be forced into wheelchairs.

**Figure 5.** The cane’s effect in reducing the load on the contralateral limb is suggested by this set of schematic drawings. Note in (B) and (C) the manner in which the hip abductor muscle normally plays an important role in controlling the descent of the pelvis on the unsupported side. When a cane supports one end of the pelvic lever (via the arm and the structure of the trunk) the cane’s support allows the pelvis to be maintained essentially horizontal even if the hip abductor contribution is small or
An additional benefit of cane use is improved stability. The size of the effective base of support is increased during single-limb-support (Fig. 5) and double-limb-support as well (Fig. 6). As long as the action line of the center of gravity of the ambulator is within the bounds of the base area, the ambulator is geometrically stable. Clinically, this increase in stability is particularly useful to persons with diminished ability to recover balance, and to disabled persons negotiating inclines or moving about in high winds.

![Diagram showing the effect of cane use on the size of the effective base of support.]

**FIGURE 6.**—The effect of cane use on the size of the effective base of support.

It is possible that the braking and accelerating impulses imparted to the cane are fringe benefits and that the basic function of the cane use of these individual men with hip pain lies elsewhere. In attempting to relieve pain or increase stability, the user may inadvertently apply the cane to the floor at angles near the beginning and end of the stance phase which contribute to restraint and propulsion. It can only be said that those suffering hip pain do use a cane in a manner suitable to aid restraint and propulsion, and that the personal cost in terms of muscular effort is high.
CONCLUSIONS

Nine experienced cane users who had hip pain resulting from degenerative joint disease or avascular necrosis were tested to determine the braking and accelerative impulses imparted to their canes during walking. It was learned that:

1. The average cane braking impulse developed in gait was 0.66 lbf-sec, or 14 percent of the equivalent value generated by a normal lower limb (14 percent of a limbsworth). Braking impulse values ranged from 0.11 lbf-sec to 2.28 lbf-sec.

2. The average cane acceleration impulse developed in gait was 1.41 lbf-sec, or 31 percent of the equivalent value generated by a normal lower limb. Accelerating impulse values ranged from 0.59 lbf-sec to 3.86 lbf-sec, and the accelerating impulses were higher than the braking impulses in eight of the nine subjects.

REFERENCES

6. Eberhart, H.D., et al.: Fundamental Studies of Human Locomotion and Other Information Relating to Design of Artificial Limbs, University of California, Berkeley, 1, Fig. 8-11, (Floor Reactions on the Foot—Northrop Footplates) 1947.