Hip Abductor Muscle Activity as Subjects With Hip Prostheses Walk With Different Methods of Using a Cane

Background and Purpose. Using a cane held contralateral to a prosthetic hip is presumed to be an effective way to reduce the demands on the hip abductor (HA) muscles and, therefore, the forces on the implant. In this study, surface electromyographic (EMG) activity was measured from the HA muscles to test this notion. Subjects. Twenty-four active subjects (9 female, 15 male) with unilateral prosthetic hips were tested. The subjects, aged 40 to 86 years ($\bar{X}=63.3$, SD=10.7), were not regular cane users. Methods. Surface EMG activity and cane force were analyzed while the subjects walked with the cane held (1) contralateral to the prosthesis (CL-CANE), (2) ipsilateral to the prosthesis (IL-CANE), and (3) contralateral to the prosthesis with instructions for the subject to push with a “near-maximal effort” (CL-CANE+). Results. Only the following conditions showed a change in HA muscle EMG activity as compared with not using a cane: CL-CANE=$-31.1\%$, CL-CANE+=$-42.3\%$. Conclusion and Discussion. Holding the cane contralateral to the prosthetic hip appears to be an effective method of reducing demands on the HA muscles. [Neumann DA. Hip abductor muscle activity as subjects with hip prostheses walk with different methods of using a cane. Phys Ther. 1998;78:490–501.]

Key Words: Cane, Electromyography, Hip abductors, Hip joint, Prosthesis.

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Several studies have made direct in vivo force and pressure measurements on an instrumented hip prosthesis in humans. These studies were designed to determine the extent to which exercise, gait, and other activities of daily living generate forces on the prosthetic hip. Reducing this load, or force, may reduce the risk of premature loosening of the device. Loosening of the femoral or acetabular component of the prosthetic hip is a major postoperative problem.

Forces on the prosthetic or normal hip are caused primarily by the effects of acceleration of body weight over the joint and the action of muscles, most notably the hip abductor (HA) group. Using a cane held in the hand contralateral to a particular hip has been advocated as a practical and effective means of reducing the forces on the hip. The data supporting this claim were collected primarily by way of indirect laboratory measurements made on subjects with hip replacement and subjects with arthritis of the hip. The use of the cane is based primarily on the premise that using the cane reduces the force demands on the overlying HA muscles and, therefore, reduces forces across the joint.

The relationship between cane force, the side of its application (ie, contralateral or ipsilateral to a given hip), and the forces generated by the HA muscles while walking has not been reported. The purpose of my study was to use surface electromyography (EMG) to indirectly assess the demands placed on the HA muscles as persons with a prosthetic hip walked while using a cane.

The HA muscles provide the primary frontal-plane stability to the hip during the mid-stance phase of walking (ie, during single-limb support). The gluteus medius muscle has the greatest mechanical advantage for this action. Other abductors include the gluteus minimus muscle, the tensor fasciae latae muscle, and the anterior fibers of the gluteus maximus muscle. The HA muscles provide frontal-plane stability to the hip by producing a torque that counteracts the torque produced by body weight (Fig. 1). The force involved with this torque is transferred across the joint and, in the case of a prosthetic hip, across the implant. Because the length of the moment arm of the HA muscles is only about one half the length of the moment arm used by body weight (D versus D1 in Fig. 1), the HA muscles must produce a force about twice body weight to hold the pelvis stable in the frontal plane. During every mid-stance phase of the gait cycle, this HA muscle force—plus the pull of body weight—is passed on to the hip joint. What is not depicted in the static model in Figure 1 is the additional force required by the HA muscles to decelerate the frontal-plane acceleration of the pelvis on the femoral head. In total, a downward force of about 3 to 3.5 times body weight is exerted on the femoral head during each gait cycle. The femoral head, in turn, must generate an equivalent force against the acetabulum. This force is

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Figure 1.
(A) Hip abductor force (HAF) from the right hip abductor muscles produces a torque necessary for frontal-plane stability of the pelvis during right single-limb support. Assuming static equilibrium, rotary stability is achieved when the counterclockwise torque equals the clockwise torque. The counterclockwise torque equals hip abductor force (HAF) times its moment arm (D), and the clockwise torque equals body weight (BW) times its moment arm (D'). (B) A first-class lever seesaw model simplifies the model shown in Figure 1A. Because the moment arm used by BW (D') is about twice the length of the moment arm used by the muscle force (D), the HAF must be at least twice the force of BW. For structural stability at the hip, the prosthetic hip “reaction force” (PHRF) must be directed upward at a magnitude equal to the sum of the HAF and BW. (Reprinted and modified from Neumann DA. Biomechanical analysis of selected principles of hip joint protection. Arthritis Care and Research. 1989;2:146–155, with permission of the American College of Rheumatology.)

Based on the reasoning presented in Figure 1, the force that crosses the prosthetic hip is directly related to the force produced by the HA muscles. In theory, applying a force to a cane held in the hand contralateral to the prosthetic hip produces a torque about the implant in the same rotary direction as that normally required by the HA muscles. Reducing the demand on the HA muscles should, in theory, reduce a significant portion of the muscle-produced force at the prosthetic hip.

Neumann and colleagues have previously used surface EMG to indirectly assess the functional demands on the HA muscles while walking and carrying loads. In the present study, I used a similar approach and rationale but focused on the response of the HA muscle to the application of a cane force. The only other study that could be found that measured surface EMG activity of the HA muscles while using a cane was performed by Vargo et al. In their study, subjects with normal hips applied a predetermined cane force while standing in one-legged or two-legged stance. Because the measurements were not made while walking, it is my opinion that the relationship between cane force and HA muscle EMG activity has not been adequately documented. In the present study, subjects had unilateral prosthetic hips, and I collected EMG data from the HA muscles while the subjects walked and applied a self-selected cane force. This research design tested a hypothesis that applying a cane force by the hand contralateral to a prosthetic hip reduces the demands placed on the HA muscle. In order to more thoroughly understand the overall biomechanics of using a cane, subjects walked while applying two different “effort levels” of cane force and, in addition, used the cane in the hand contralateral and ipsilateral to the prosthetic hip.
The primary measurements made in this study were normalized surface EMG activity produced by the HA muscles and the force applied through the cane. A secondary and less important measurement was normalized EMG activity from the triceps surae muscles. The triceps surae muscle data served as a general indicator of the relative effort exerted by the subjects during different methods of using the cane. This study focused only on the data produced during the mid-stance phase of walking. Based on the model described in Figure 1 and other research, changes in HA muscle EMG activity were assumed to reflect a similar relative change in muscle force produced across the prosthetic hip. Because the mathematical relationship between EMG activity and muscle force is not known for the HA muscles, no attempt was made to use EMG activity as an absolute measurement of force.

Method

Subject Selection Process

Twenty-four relatively active persons with one hip prosthesis were selected for the study. At the time of the study, they lived in or near a moderately large metropolitan midwestern city in the United States. All subjects were paid for their time and received consultation regarding their current exercise program and suggestions on ways to minimize stress on their prosthetic hip.

Subjects selected for this study had to meet the following criteria. There must be only one prosthetic hip, and the prosthetic hip must not be the result of a revised procedure. The hip replacement must be secondary to osteoarthritis. Persons with a hip implant due to rheumatoid disease, avascular necrosis, or congenital dysplasia were not selected. Furthermore, the prosthetic hip must be the only joint prosthesis in the subject’s body, and all other joints must be relatively pain-free. Several subjects selected for the study stated that they experienced minor joint pain, but the pain was not severe enough to be considered disabling.

All subjects selected for this study declared that they were in good health and independent in all activities of daily living. No subject reported having respiratory disease, heart or vascular disease, or severe diabetes. Subjects did not normally use an assistive device for walking for distances less than 0.8 km (½ mile). This specification was necessary because the EMG normalization process used for the HA muscles required that subjects walk multiple trials without the use of cane. Earlier unpublished work showed that subjects who were regular cane users could not adequately perform this part of the experiment. Finally, subjects selected for this study did not require any special footwear or orthosis for walking.

Subject Profile

Nine women and 15 men were selected for this study. All subjects signed consent forms as required by the Human Subjects Review Committee of Marquette University. Subjects ranged in age from 40 to 86 years (X=63.3, SD=10.7), in weight from 498.2 to 1,085.3 N* (X=757.5, SD=166.8), and in height from 1.52 to 1.91 m (X=1.72, SD=0.1). Twelve subjects had the prosthetic hip on their right side, and 12 subjects had the prosthetic hip on their left side.

I performed a physical examination on each subject and gave a questionnaire to each subject prior to his or her acceptance into the study. During the examination, I checked for weakness in major muscle groups of the lower extremity (through manual resistance), hip instability, gait abnormality, pain, or other conditions that may affect subject safety.

Figure 2.
The cane is shown with force transducer built into the lower shaft.

* 4.448 N=1 lb.
their left side. The time since surgery ranged from 5 to 96 months (X=24.9, SD=21.4).

**Instrumentation**

The EMG instrumentation used in this study has been described previously. The EMG unit consisted of surface on-site electrodes, a ground electrode, an oscilloscope, a signal conditioning unit, a personal computer and analog-to-digital convertor, and software for data collection and data reduction. Raw bipolar EMG data were processed using the root-mean-square (RMS) method to produce a linear envelope, or average voltage, over a specified time. The time constant used for the RMS processing was 55 milliseconds. The sampling rate of the processed EMG data was 100 times per second. A calibrated electronic force transducer was mounted in series in the stem (or shaft) of a standard, aluminum adjustable cane from the rubber tip (Fig. 2). The force transducer recorded the axial compression force produced through the long axis of the cane. The instrumented cane was calibrated by loading weights directly through the stem of the cane. The voltage-load calibration curve was incorporated into the computer software. Before and after each experiment, the calibration of the instrumented cane was tested by applying a known weight through the stem of the cane. The voltage-load calibration curve was incorporated into the computer software. Before and after each experiment, the calibration of the instrumented cane was tested by applying a known weight through the stem of the cane. The output voltage from the transducer was adjusted to ignore the actual weight of the cane.

Subjects wore footswitches attached to rubber galoshes placed over their shoes. The footswitches produced voltages that associated all data with a particular phase of gait. Two on-off footswitch closures defined the mid-stance phase of gait as the time interval between the instant of footflat and just prior to heel-off. Electromyographic activity, cane force, and footswitch voltages traveled between subject and signal processor and computer via a single 12.2-m (40-ft) cable.

**Procedure**

**Pre-experimental protocol.** Subjects were taken to a room for the application of the EMG electrodes, ground plate, and rubber galoshes. The skin over both right and left posterolateral gluteal regions and posterior arms was thoroughly cleaned with alcohol. The EMG electrodes were then placed on the skin superficial to both right and left bellies of the gluteus medius muscle as described in earlier work.

Electrode placement was verified by palpation of the gluteus medius muscle during isometric contraction and by observation of the raw EMG signal as the subject stood in single-limb support on the side of the active muscle. The EMG electrodes were also placed on the posterior surface of both arms, at a distance halfway between the acromion and the olecranon process. The ground electrode was placed over the anteromedial aspect of the tibia on the side of the prosthetic hip.

The height of the cane was adjusted to the appropriate height of each subject in the following manner. Subjects stood with a relaxed posture with the tip of the cane placed on the floor, 10.2 cm (4 in) lateral to the small toe. The height of the cane was then adjusted so that the elbow angle measured 30 degrees of flexion.

Subjects were taught to walk at a relatively constant self-selected walking speed on an indoor, hard-surface walkway. They were told to walk at their “natural speed” while using a cane held in the hand opposite their prosthetic hip. After at least 3 minutes of walking, each subject’s average walking speed was determined by a stopwatch to the nearest 10th of a second (X=0.82 m/s, SD=0.09, range=0.61-1.0, for all 24 subjects). Each subject’s average walking speed was determined over three trials of walking a 10-m distance. Subjects repeated all subsequent walking trials at a speed within 10% of their own self-selected target speed.

**Experimental protocol.** Subjects practiced walking in a natural manner with the instrumentation in place but without using the cane. Before the start of the experiment, a pre-experimental EMG baseline was established for each subject. The HA muscle EMG baseline was determined by averaging the sampled HA muscle EMG voltage data produced during the mid-stance phase of walking at the subjects’ self-selected walking speed. During this part of the experiment, each subject walked without the use of a cane. For each walking trial, the sampling of EMG data began as the subject walked across a 2-m mark on the walkway and continued for 10 seconds. Subjects were instructed to stop walking after the 10-second period. An EMG baseline voltage was determined by averaging these data across the four walking trials.

A triceps surae muscle EMG baseline was determined by averaging the EMG voltage produced from the triceps surae muscles during a maximal-effort reference contraction. Subjects stood holding one cane in a manner similar to that described for adjusting cane height. Each subject was then told to push down on the cane as hard as possible for about 7 seconds. At 2 seconds after the onset of muscle activity, EMG data were collected for the next consecutive 3 seconds. Four trials were repeated for each arm, with a 30-second rest between trials. The triceps surae muscle EMG baseline voltage was deter-
mined by averaging these data across the four trials.

During the experiments, subjects walked while data were collected during three different cane conditions. In the first condition, subjects held the cane in the hand contralateral to the prosthetic hip (referred to as the CL-CANE condition). Subjects were instructed to place the cane on the floor at the same time the foot of the "operated side" was on the ground. Subjects were instructed to push on the cane with a "moderate but comfortable" force. The second condition consisted of the subjects using the cane held in the hand ipsilateral to the prosthetic hip (referred to as the IL-CANE condition). Subjects were instructed to place the cane on the floor at the same time the foot of the "operated side" was on the ground. As with the CL-CANE condition, subjects were instructed to push on the cane with a "moderate but comfortable" force. The order of performing the CL-CANE and IL-CANE conditions was random for each subject.

After the data were collected for the first two conditions, data were collected for the third condition. This condition was similar to the CL-CANE condition, except that subjects were told to push on the cane with a "near-maximal effort." This third condition was referred to as the CL-CANE+ condition, with the plus sign designating instructions to generate a near-maximal effort force. This final experimental condition was used to determine how different magnitudes of cane force, when produced by the contralateral hand, affect HA muscle EMG activity. Of the three conditions, it is my belief that the CL-CANE condition most closely reflects the instructions given by physical therapists to persons who use a cane following a prosthetic hip implant.

For each cane condition, subjects were allowed several minutes to practice using the cane. Subjects were allowed to repeat any trial that the subject or the experimenter felt was not performed as previously defined.

The method of collecting data during the experimental walking trials was similar to the method described for the pre-experimental phase of the experiment. One difference, however, was that two walking trials of data were collected for each of the three cane conditions. This experimental design provided data on approximately 16 gait cycles per subject per cane condition. The HA muscle EMG voltages produced as subjects used a cane were normalized to a percentage of the pre-experimental voltage baseline. This normalized HA muscle EMG value was expressed as a percentage of the baseline activity (%EMG).

A 90-second rest was permitted between walking trials. Data were accepted for analysis only after the target walking speed was confirmed and a typical footswitch pattern was displayed on the computer screen.

Reliability Assessment of HA Muscle EMG Data
After data were collected for the three cane conditions, each subject was asked to establish a post-experimental EMG baseline by repeating the pre-experimental walking trials. To determine the intrasubject reliability of the HA muscle EMG measurements, a comparison was performed between the grand mean EMG activity (in millivolts) produced in the pre-experimental walking trials and that produced in the post-experimental walking trials. Approximately 3 hours separated these two measurements. Each grand mean was calculated by averaging all 24 subjects' EMG voltage from the side of the prosthetic hip during the mid-stance phase. The pre-experimental no-load EMG voltage mean was 147.4 mV, and the post-experimental no-load EMG mean was 141.3 mV. I considered this 4% decrease in EMG baseline as insignificant. An intraclass correlation coefficient (ICC[1,k]) of .99 was calculated for the association between the pre-experimental and post-experimental no-load EMG data (P < .0001).\(^{21,22}\)

Data Analysis
The independent variable, "cane condition," consisted of CL-CANE, CL-CANE+, and IL-CANE conditions. The primary dependent variables associated with each cane condition were HA muscle %EMG on the side of the prosthetic hip and cane force (averaged over the mid-stance phase). A secondary dependent measurement was triceps surae muscle EMG activity (expressed as a percentage of a maximal-effort reference contraction [%MRC]). These three values were expressed as a grand mean, based on approximately 16 mid-stance gait cycles per subject, averaged over all 24 subjects. An analysis of variance (ANOVA) with a repeated-measures design was performed using each dependent variable. A separate post hoc test was performed on each dependent variable. Each dependent variable was compared against each of the three cane conditions; the HA muscle %EMG was also compared against zero (i.e., the pre-experimental EMG baseline, when no cane was used while walking). For these tests, a multiple t test with Bonferroni adjustments was used.\(^{17,22}\) These adjustments maintained the a priori alpha level by dividing .05 by the number of preplanned comparisons. Comparisons between means were only assumed to be different when the probability value for each comparison was less than the adjusted alpha level.
Table. Descriptive Statistics for Hip Abductor (HA) Muscle %EMG, a Cane Force, and Triceps Surae Muscle EMGb for Three Cane Conditionsc (N=24)

<table>
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<th></th>
<th>X</th>
<th>SD</th>
<th>Minimum</th>
<th>Maximum</th>
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<td><strong>HA Muscle %EMG</strong></td>
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<tr>
<td>CL-CANE</td>
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<td>15.6</td>
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<td>CL-CANE+</td>
<td>-42.3</td>
<td>16.2</td>
<td>-65.7</td>
<td>-9.0</td>
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<td>IL-CANE</td>
<td>3.8</td>
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<td>-39.0</td>
<td>44.0</td>
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<tr>
<th><strong>Cane force (N)</strong> (force in pounds is in parentheses)</th>
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<tbody>
<tr>
<td>CL-CANE</td>
<td>76.1 (17.1)</td>
<td>42.3 (9.5)</td>
<td>7.6 (1.7)</td>
<td>165.5 (37.2)</td>
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<tr>
<td>CL-CANE+</td>
<td>149.9 (33.7)</td>
<td>60.5 (13.6)</td>
<td>52.9 (11.9)</td>
<td>310.5 (69.8)</td>
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<tr>
<td>IL-CANE</td>
<td>58.7 (13.2)</td>
<td>35.1 (7.9)</td>
<td>8.0 (1.9)</td>
<td>138.8 (31.2)</td>
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<th><strong>Triceps surae muscle EMG</strong></th>
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<td>50.6</td>
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<tr>
<td>CL-CANE+</td>
<td>154.7</td>
<td>64.5</td>
<td>66.0</td>
<td>298.0</td>
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<tr>
<td>IL-CANE</td>
<td>91.5</td>
<td>62.2</td>
<td>17.6</td>
<td>272.0</td>
</tr>
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</table>

a %EMG=percentage of electromyographic voltage produced during walking without a cane. Negative values indicate EMG less than that produced while walking without a cane.
b EMG=percentage of electromyographic voltage produced during a maximal-effort reference contraction.
c CL-CANE=cane held contralateral to the prosthesis, CL-CANE+=cane held contralateral to the prosthesis with instructions to push with a "moderate but comfortable" effort, IL-CANE=cane held ipsilateral to the prosthesis.

Results

The descriptive statistics for HA muscle %EMG, cane force, and triceps surae muscle EMG activity for all three cane conditions are displayed in the Table.

An ANOVA on the mean HA muscle %EMG showed a main effect for cane condition (F=40.62, P<.0001). The %EMG means for each condition are shown in Figure 3. A negative %EMG value indicates that the EMG voltage was less than that produced while walking without a cane.

Both the CL-CANE and CL-CANE+ conditions produced HA muscle %EMG values that were different from 0% (ie, the EMG produced while walking without a cane). The mean HA muscle %EMG for the IL-CANE condition, however, was equivalent to zero. Of particular note was that the mean HA muscle %EMG produced during the CL-CANE condition (ie, -31.1%) was different from the %EMG produced during the CL-CANE+ condition (ie, -42.3 %).

An ANOVA on the average cane force and triceps surae muscle EMG activity (%MRC) showed a main effect for cane condition (F=37.97, P<.0001 and F=13.39, P<.0001, respectively). These data are plotted against cane condition in Figure 4. The average cane force and triceps surae muscle EMG activity produced during the CL-CANE+ condition were different from those produced during both CL-CANE and IL-CANE conditions. The average cane force and triceps surae muscle EMG activity, however, were not different between the CL-CANE and IL-CANE conditions.

Discussion

Using the Cane in the Hand Contralateral to the Prosthetic Hip

Applying a "moderate but comfortable" cane force effort. In the CL-CANE condition, subjects were instructed to push on the cane with a "moderate but comfortable" effort. On average, subjects responded to this instruction by producing an average cane force of 76.1 N (17.1 lb). This force is equal to 10% of the subjects' average body weight, only slightly lower than that reported for other studies that tested a similar population. Applying a "moderate but comfortable" cane force by the hand contralateral to the prosthetic hip is an effective method of reducing the demands on the HA muscles.

The general "inverse" relationship between cane force and HA muscle %EMG can be understood by the use of a simplified frontal-plane model (Fig. 5A). Acting through the moment arm (D₂), the application of a cane force (CF) by the left hand, for example, produces a frontal-plane torque (CF × D₂) about the prosthetic hip. The counterclockwise rotation of this cane-generated torque is in the same rotary direction as that ordinarily produced by the HA muscles (ie, HAF × D). The cane force and HA muscles, in effect, function as a "force couple" that oppose the external torque produced by body weight (BW × D).

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The amount that the cane theoretically unloads the contralateral hip can be estimated by performing calculations based on an assumption of static equilibrium during the mid-stance phase of walking (Fig. 5B). The calculations are based, in part, on the subjects' mean body weight and the average cane force applied during the mid-stance phase of the CL-CANE condition. As shown by the torque equilibrium equation, 660.6 N (148.5 lb) of hip abductor force are needed for frontal-plane stability when applying a cane force of 76.1 N (17.1 lb). Based on the force equilibrium equation depicted in the figure, this amount of abductor force would result in a reaction force at the prosthetic hip of 1,228.4 N (276.2 lb). Using the same data and equations supplied in Figure 5B but eliminating the cane variables (CF and Dc), calculations show that 1,910.4 N (429.5 lb) of prosthetic hip reaction force would result when not using a cane. Based on this model, therefore, the CL-CANE condition theoretically reduces the prosthetic hip reaction force by 35%. 

Figure 3.
Plot showing the mean hip abductor percentage of electromyographic voltage (%EMG) (expressed as a percentage of baseline EMG voltage) as subjects with a prosthetic hip walked while using a cane in three conditions: CL-CANE (cane held in the hand contralateral to the prosthetic hip), CL-CANE+ (similar to CL-CANE condition, but subjects were instructed to push with a "near-maximal" cane force effort), and IL-CANE (cane held in the hand ipsilateral to the prosthetic hip). Data are shown as an average of all subjects (N=24) during the mid-stance phase of the walking cycle. Negative %EMG indicates electromyographic voltages less than that produced while walking without a cane. Brackets about the means indicate standard error. Asterisk (*) indicates means were statistically different from the zero baseline. (The small figures with the cane assume that the right hip is the prosthetic hip.)
In 1956, Blount published one of the first articles describing this biomechanical aspect of the cane, basing his argument on static equilibrium principles. More recently, Brand and Crowninshield used a dynamic model to test the efficacy of using the cane in a group of patients with hip disability. They reported a 40% reduction in peak hip "contact force" when subjects used a cane in the hand opposite the affected hip while walking at a speed of 0.44 m/s. This reduction in hip force is similar to the 35% reduction estimated by using the static model shown in Figure 5. Interestingly, the HA muscle %EMG measured in this study during the CL-CANE condition was reduced by a similar amount (ie, 31%). This similarity suggests a reasonably close association between the reduction in HA muscle %EMG and the theoretical reduction in forces at the prosthetic hip when using the cane. The actual relationship between these variables, however, is not known.

Applying a "near-maximal" cane force effort. In the CL-CANE+ condition, subjects were instructed to push on the cane with a "near-maximal effort." As depicted in Figure 4, subjects nearly doubled the average cane force applied during the CL-CANE condition, from 10% to 19.8% of the subjects' average body weight (76.1 N [17.1 lb] to 149.9 N [33.7 lb]). This larger cane force was reflected by an increase in triceps surae muscle EMG activity (from 85.7 %MRC during the CL-CANE condition to 154.7 %MRC during the CL-CANE+ condition). Pushing with a substantially greater cane force resulted, on average, in a further reduction of HA muscle %EMG, to 42.3% below that produced when not using a cane (Fig. 3). Note that the average triceps surae muscle EMG activity during the CL-CANE+ condition greatly exceeded the 100% level (ie, the EMG activity produced during the "maximal effort" reference contraction). Changes in elbow angle or leaning toward the
cane while walking may partially account for this higher value.

The model depicted in Figure 5 predicts that increasing the cane force would cause a further reduction in HA muscle %EMG. Figure 6 shows the HA muscle %EMG versus cane force data for all subjects during both CL-CANE and CL-CANE+ conditions. When considered over both contralateral cane force conditions, the two variables were correlated in a negative direction (ie, greater cane forces were associated with less HA muscle %EMG, and vice versa). The strength of this association was low ($r = -0.44$). This low correlation implies that other unknown factors, in addition to cane force, are involved in the reduction of %EMG.

Based on the data shown in Figure 6, increasing the amount of cane force (with the cane held contralateral to the prosthetic hip) will, on average, decrease the demands on a muscle group that is primarily responsible for controlling forces at the hip. I do not believe that these results should imply that clinicians should advise patients with prosthetic hips to apply a cane force that exceeds their comfort level. In my study, the reduction in HA muscle %EMG per unit of cane force was different for the CL-CANE and CL-CANE+ conditions. Doubling the cane force from 10% to nearly 20% of body weight only reduced the HA muscle %EMG from about −31% to −42%. Pushing with a cane force of nearly 20% of body weight may be biomechanically less "efficient" in reducing the demands on the HA muscles. Without knowing whether the center of body mass changed with the application of greater cane force, this issue cannot be adequately addressed. Whether the decreased reduction in %EMG per unit of cane force translates to a decreased reduction in hip force per unit of cane force also cannot be determined from this study. Even assuming that HA muscle %EMG reflects the actual forces at the hip, the issue of metabolic efficiency arises. At what point does the increased physiologic work required to generate a larger cane force outweigh the possible benefit of protecting the contralateral prosthetic hip? This question should be addressed before valid advice can be given on how much force should be applied to

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**Figure 5.**

(A) Diagram showing a balance of frontal-plane torques about the right prosthetic hip while in single-limb support and using a cane held by the hand contralateral to prosthetic hip. Assuming static equilibrium, the sum of the clockwise torque produced by body weight (BW) (dashed circle) equals the counterclockwise torques produced by the hip abductor force (HAF) and cane force (CF) (solid circles). D_{BW} = moment arm used by BW, D_{HAF} = moment arm used by the cane force. The prosthetic hip reaction force (PHRF) is equal in magnitude to the inferior-directed forces described above. The force vectors are not drawn to scale. (B) The data in the box are based on laboratory measurements and average body weight and cane force produced in the CL-CANE condition. Additional moment arms are based on other research.24,25 Assuming static frontal-plane balance of torque about the right prosthetic hip, relatively simple calculations estimate the approximate HAF and PHFR during use of the cane. (Reprinted and modified from Neumann DA. Hip abductor muscle activity in persons with a hip prosthesis while walking carrying loads in one hand. Phys Ther. 1996;76:1320-1330, with permission of the American Physical Therapy Association.)
maximize hip protection. Research should include measurements on the kinematics of walking, oxygen consumption, and, if available, the actual forces delivered to the prosthetic hip.

Using the Cane in the Hand Ipsilateral to the Prosthetic Hip

Based on the literature reviewed and the experience of the author, the IL-CANE method is not the standard method of advising a person to walk with a cane following a hip replacement. Returning to the model in Figure 5, it is apparent that applying a cane force by the subject’s right hand would cause a clockwise torque about the prosthetic hip. This cane-generated torque would be in opposite directions to that naturally produced by the right HA muscles. Theoretically, therefore, using the cane in this manner would increase the force demands on the HA muscles and presumably on the prosthetic hip. In my study, however, the IL-CANE condition produced a mean HA muscle %EMG of +3.8% (Fig. 3). This magnitude was statistically equivalent to 0% (ie, the amount of normalized EMG activity produced when not using a cane). Why the IL-CANE condition did not produce HA muscle %EMG greater than zero cannot be explained with certainty. A gait pattern with the arm (and cane) “in phase” with the sagittal-plane kinematics of the ipsilateral lower limb is not a natural method of walking. Although not measured, it appeared that some subjects in this study leaned their upper body slightly toward the side of cane application. If indeed true, this lean toward the side of the prosthetic hip may reduce some of the demand on the HA muscles by reducing the length of the moment arm used by body weight (See D, in Fig. 1).

In summary, although the IL-CANE condition did not increase the demands on the HA muscles, this method of using the cane did not reduce HA muscle %EMG below that produced while walking without a cane. The IL-CANE condition, therefore, is not considered an effective method for reducing the demands on the HA muscles over the prosthetic hip.

Limitations of This Study

The model used in Figure 5 was based on the assumption that static equilibrium exists over the prosthetic hip during the midstance phase of walking. For simplicity, the model did not include dynamic variables associated with the mid-stance phase, forces and torques generated outside of the frontal plane, or changes in the center of mass while walking. Furthermore, the model assumed that all forces acted in the vertical direction. This model, therefore, contains error when estimating the absolute force and torque magnitudes. The model does, however, allow a basic framework for understanding the approximate relative magnitudes of hip abductor-generated forces at the prosthetic hip when a person uses a cane as described by the CL-CANE condition.

Subjects in this study used a cane for varying times during their postsurgical physical rehabilitation period. At the time of data collection, however, subjects no longer needed a cane for walking distances less than 800 m. It is possible that the results of this study may be different for persons who require a cane for shorter walking distances.

Final Comment and Conclusions

The decision on when or under what circumstances a cane should be used to protect the prosthetic hip is
based on several factors, including but not limited to method of surgical fixation; time after surgery; presence of osteoporosis; history of failed procedures; and age, activity level, and mental status of the patient. A discussion of these factors was beyond the scope of this report. Assuming, however, that a cane is warranted for whatever reason, this study supports the principle that the cane be used in the hand contralateral to the prosthetic hip. This method of cane use is an effective method of reducing demands on the HA muscles and presumably on the prosthetic hip. When instructed to push on the cane with a "moderate but comfortable" cane force (ie, CL-CANE condition), subjects produced an average cane force equal to 10% of the subjects' average body weight (76.1 N [17.1 lb]). This level of effort produced a 31% reduction in HA muscle %EMG below that generated while not using a cane.

When instructed to push on the cane with a "near-maximal effort" (ie, during the CL-CANE+ condition), subjects doubled their average cane force to 19.8% of the subjects' average body weight (149.9 N [33.7 lb]). This more strenuous effort produced, on average, a 42.3% reduction in HA muscle %EMG below that generated while not using a cane. The degree to which these reductions of EMG activity actually reflect the presumed reduction in forces on the prosthetic hip cannot be determined from this study.

Using the cane in the hand ipsilateral to the prosthetic hip resulted in a 3.8% increase in HA muscle %EMG above that generated while not using a cane. This value was not statistically different from the amount of EMG activity produced when not using a cane. Using the cane in this fashion is not considered an effective method of reducing the demands on the HA muscles.

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