Electromyographic Analysis of Hip Abductor Musculature in Healthy Right-handed Persons

The right hip abductor (HA) musculature has been reported to demonstrate stretch weakness attributable to chronic elongation imposed by standing posture common to right-handed healthy individuals. Research has shown side-specific differences in maximal isometric torque-hip abduction angle curves for the HA muscles in a similar population. The purposes of this study were to assess isometric hip abduction torque in a sample of 40 healthy right-handed persons and to compare side differences in the HA muscles' neural drive index across a wide range of hip abduction angles. The neural drive index was defined as the magnitude of the normalized surface electromyographic activity divided by the absolute magnitude of a submaximal (20% of maximal) isometric HA muscle torque. Data were collected while subjects were positioned supine with trunk and pelvis stabilized. Statistical analysis of the data did not support earlier research on right HA muscle stretch-weakness; however, trends in the data suggest the need for further research. We also discuss the functional role of the passive noncontractile elements of the HA musculature.

Kendall and associates have described the concept of "stretch weakness" of muscle groups following elongation imposed by asymmetry in standing posture. One such muscle group is the right hip abductor (HA) muscles in right-handed persons, which, according to Kendall, typically remain at a relatively stretched and longer length than the left HA muscles. The predominant muscle of this muscle group is the gluteus medius muscle. Neumann et al have tested for and attempted to physiologically define the condition of right HA muscle stretch weakness. These authors were not able to demonstrate a specific weakness (ie, torque diminution) of the right HA muscles in their subjects, but did discover differences in the maximal isometric torque-angle curves across hip sides, which may partially account for Kendall's observations of the stretch-weakness phenomenon in general. This research is an extension of the work of Neumann et al with a primary purpose of exploring the concept of stretch weakness by examining contractile component activity via electromyography. A secondary purpose was to use the EMG data to observe in general the functional aspects of the HA musculature.

Background of the Study

The standing posture which Kendall and McCreary report to be typical of many right-handed persons has been depicted elsewhere. The relevant aspect of this posture is that the right hip is in slight adduction and the left hip is in slight abduction. This posture of the right hip will be referred...
to as pelvic adduction. This asymmetry in standing posture would place the right HA muscles at a slightly longer length than the left HA muscles. According to Kendall and associates, the relative elongation of the right HA muscle group, over time, predisposes to postural stretch weakness.\(^1,2,4\) Animal experiments have shown that skeletal muscles, when immobilized at lengths longer than their functional length (i.e., the length at which a muscle most often functions), show alterations in their length-tension relationship such that peak tensions are produced at the same elongated length imposed by the experimenters (Fig. 1).\(^5-8\) This alteration in the elongated muscles' resting length (i.e., the length at which a muscle generates maximal tension) creates a shift in the immobilized muscles' length-tension curve. This shift in the length-tension relationship may be an adaptive process of muscle that allows the peak tension to be available at the muscles' functional length.\(^7\) Extending this concept to persons with HA muscle length asymmetry, the reported right HA muscle stretch weakness may be a secondary adaptive shift in the torque-angle curves between right and left HA muscles.\(^3,9\) The observed phenomena of right HA muscle stretch weakness may be due to the short length at which the HA muscles are manually tested,\(^1\) and the right HA muscles may produce greater force at the longer lengths of hip adduction.\(^9\) Using this logic, Neumann et al\(^3\) compared the maximal isometric torque-hip angle relationship across right and left HA muscles using right-handed persons who displayed standing postures that, according to the teachings of Kendall,\(^1,2,4\) are typical of the right-handed person. Neumann et al hypothesized that a posturally induced stretch weakness of the right HA muscles, if present, may be demonstrated in humans by their muscle torque-hip angle curves, which would be similar to the animal data displayed in Figure 1. Figure 2 shows the maximal torque-angle data generated from their human research after the data were subjected to a linear regression. The authors demonstrated that a specific weakness was not evident in the right HA muscles. There were slight, but statistically significant, differences in the torque-angle slopes across hip sides, however, that were similar to those reported in the animal experiments.\(^5-8\) The HA muscle torque-testing protocol used by Neumann et al limited the elongated length of the HA musculature to 10 degrees of adduction.\(^8\) We, therefore, speculated that a longer length equivalent to greater degrees of adduction would reveal a true “adduction shift” in the right HA muscles’ resting length. The authors did suggest that a cause-and-effect relationship may exist between the posture of right pelvic adduction and the differences in the torque-angle slopes across hip sides, possibly based on the adaptive nature of the muscles. Regardless of hip side, the maximal isometric torque potential of the HA muscles decreases as the hip joint angle approaches greater angles of abduction.\(^10,11\) At the sarcomere level, this decrease may be explained by the fact that the shortened length of the muscles may minimize the available sites for actin and myosin overlap,\(^12\) a condition often referred to as a muscle “on slack,” or a muscle exhibiting “active insufficiency.”\(^13\) We speculate that the torque differences between hip sides at the various muscle lengths reflect a posturally induced side-specific difference in the available actin and myosin overlap. Furthermore, the shift in the resting length of the right HA musculature to longer (i.e., adducted) lengths theoretically should express itself by the right and left HA muscles displaying different levels of active insufficiency when tested at equivalent muscle lengths.

![Fig. 1.](image-url) **Fig. 1.** Active length-tension curves of “control” animal soleus muscle versus soleus muscle that was immobilized for three weeks in an elongated position. Arrow indicates the length at which the immobilized muscle was held during the experiments. (Adapted from Williams and Goldspink\(^7\) and reprinted from Physical Therapy [Vol 68, page 497, 1988] with permission of the American Physical Therapy Association.)
Different levels of active insufficiency are expected because at any given hip angle in abduction, the right HA muscles would be more slackened functionally than the left HA muscles. If experiments can show that the right and left HA muscles display different levels of active insufficiency at equivalent muscle lengths, then the hypothesis that subtle postural asymmetry would result in a shift in the resting length of the HA muscles would be strengthened. This additional evidence would support the earlier study by Neumann et al., who suggested the idea of posturally induced shifts in the functional resting length of the HA muscles based on the different torque-angle curves alone.

We used a measurement called the neural drive index (NDI) in this study to determine whether side-specific differences exist in HA muscle active insufficiency for the same subjects who participated in the earlier study by Neumann et al. The rationale for using NDI as our measurement tool was based on research by Heckathorne and Childress, who showed that for the generation of submaximal isometric muscle tension, the magnitude of the surface EMG voltage varies inversely with muscle length. The increased neural drive apparent in a muscle placed at short lengths may be due to central nervous system compensation for the local active insufficiency of the shortened muscle. In the case of the HA muscles in this study, the amount of surface EMG activity (expressed as a percentage of the EMG activity produced during maximal voluntary isometric contraction [MVIC]) generated per unit of submaximal HA muscle torque should reflect relative differences in active insufficiency between HA muscle sides. We operationally defined the NDI as the magnitude of EMG activity (% MVIC) per newton-meter of a submaximal (20% of maximal) hip abduction isometric torque effort. Dividing the EMG activity (% MVIC) by the absolute torque magnitude allowed us to normalize the EMG activity per unit of torque, therefore permitting a comparison of neural drive across hip sides.

The NDI was calculated for both right and left HA muscles across a wide range of HA muscle lengths. In theory, the hip side that yields the greatest NDI when tested at equivalent muscle lengths would be considered to display the greatest relative active insufficiency.

Our null hypothesis was that the relationship between NDI and hip abduction angle (ie, muscle length) would be equivalent in both right and left HA muscles in our right-handed subjects. Our alternative hypothesis was that the NDI-hip abduction angle relationship would be dissimilar across hip sides because of the reported "adduction bias" of the resting length of the right HA muscles.

**Method**

**Subjects**

Forty healthy right-handed volunteers (20 male, 20 female) were chosen for this study from a population of college students. All subjects signed appropriate informed consent forms as required by the university’s Human Subjects Review Committee. The subjects’ average age was 26.5 years (± 5.6 years). Their average weight was 65.5 kg (± 11.5 kg), and their average height was 1.74 m (± 0.11 m). Subjects were free of musculoskeletal deformities that may have required medical treatment. Volunteers who demonstrated a leg-length discrepancy greater than 1.3 cm were not used as subjects. All subjects demonstrated an observable right pelvic adduction asymmetry during standing, as described in

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**Fig. 2.** Best-fit abductor muscle torque-angle regression lines for both hip sides. (Reprinted from Physical Therapy [Vol 68, page 502, 1988] with permission of the American Physical Therapy Association.)
earlier research. Because of the protocol used in the EMG data collection, all subjects demonstrated at least 45 degrees of passive hip abduction.

**Instrumentation**

**Instrumentation for hip abductor muscle torque measurement.** A modified version of the Iowa Force Table was used to measure the magnitude of the maximal and submaximal HA torques. This instrument was used to generate the predictive data shown in Figure 2. This HA muscle torque-testing instrument consists of a table with appropriate trunk and pelvic stabilization devices, and a calibrated electronic force transducer with amplifier and signal processor coupled to a laboratory microcomputer. The specific design of this equipment for testing and recording HA muscle torque at angles of —10, 0, 10, 20, 30, and 40 degrees of hip abduction, including reliability testing and explanation of the stabilization and alignment protocol, was reported in an earlier study. In brief, the torque-testing system provided stabilization to the subjects' trunk and pelvis while simultaneously allowing the supine subjects to produce a maximal isometric hip abduction torque, one hip at a time, at each of the six test angles. The torque-testing device collected both maximal followed by submaximal isometric HA muscle torques for each of the 12 experimental conditions (ie, six hip angles × two hip sides). Measurement of the maximal torques was required for the normalization of the EMG signal. The 20% submaximal torque was required to calculate the NDIs. The value of 20% was chosen because we desired the NDI to approximate the magnitude of neural drive that reflected a nonfatiguing functional task, such as the HA muscle EMG activity generated during the stance phase of gait. An electronic submaximal torque feedback system was designed to provide subjects with a method for producing a level of HA muscle torque that was 20% of the torque produced by maximal effort for each particular experimental condition. The system allowed the analog force signal from the amplifier to be amplified at gains of either one or five. At a gain of one, the experimenter marked the voltage output produced by a subject's maximal torque production with an indicator located on the face of the voltage meter. The gain to the voltage meter was then changed to five, allowing the subject to produce a 20% submaximal torque by matching the deflection of the needle on the voltage meter to that previously set by the indicator. Pilot work demonstrated that subjects could consistently produce submaximal torques that were within 1% of the actual targeted submaximal torque.

**Instrumentation for electromyographic measurement.** The EMG instrumentation and specifications used in this study have been thoroughly described in other articles. In brief, the instrumentation consisted of two surface electrode assemblies in addition to a ground electrode, the signal-conditioning and amplification system, a four-channel oscilloscope, and an interactive laboratory microcomputer. The raw signal was processed by using the root-mean-square (RMS) method, which produced a linear envelope of the voltage, or "moving average" over time of the myoelectric signal. The time constant used for the RMS signal processing was 55 msec. The quality of the raw EMG signal was monitored throughout the entire experiment on the oscilloscope. The RMS EMG signals from both right and left HA muscles were sampled at 100 times per second by the microcomputer.

**Data Collection**

Data collection consisted of recording the NDI during a 20% submaximal isometric hip abduction torque effort for both right and left HA muscles across the angles of −10, 0, 10, 20, 30, and 40 degrees. Subjects were briefed on the experimental protocol and then read and signed information and consent forms. Following the physical assessment (eg, postural assessment, recording of weight and height), subjects were instructed to perform 5 to 10 minutes of supervised hamstring and hip adductor muscle stretching and light isometric (static) exercise. The skin over the region of the midbelly of the gluteus medius muscle was cleaned with alcohol. The electrode assemblies were filled with conductive electrode gel and applied to the skin at right angles to the general fiber direction of the gluteus medius muscle. The following protocol was followed for electrode placement. The anterior superior iliac spine (ASIS), the posterior superior iliac spine (PSIS), and the apex of the greater trochanter were identified and marked by grease pencil. Next, a mark was made at the superior margin of the iliac crest at a location one half the distance between the ASIS and PSIS. The electrode was placed at a point one half the distance between the greater trochanter and the mark made on the iliac crest. Cadaver dissection verified that this location was superficial to the midbelly of the gluteus medius muscle. The ground electrode was placed on the anterolateral aspect of the right proximal tibia. After verifying an artifact-free EMG signal, subjects were positioned supine on the torque-testing table and the stabilization devices were secured. A mobile skateboard was used to support the subject's heel so that the test limb remained parallel with the surface of the table. To provide familiarization and warm-up for actual data collection, subjects practiced producing maximal and 20% submaximal torque efforts at two randomly determined test angles for each hip. Subjects were allowed multiple trials to learn the psychomotor skill of the maximal and 20% submaximal effort HA muscle torque contractions. Subjects assumed about 25 degrees of hip external rotation during all HA muscle torque efforts.
Because the primary goal of this experiment was to compare NDIs between sides at equivalent angles, all within-angle trials were kept as close in time as possible. Therefore, the actual experiment consisted of six parts, one for each test angle. A random ordering of the six test angles was determined for each subject before the experiment. The ordering of the hip used first for testing was also randomized within each test angle. One half of the subjects' side-×-angle conditions were performed with the right hip first and the other half with the left hip first.

To begin the first part of the actual experiment, the subject's hip was passively abducted with the aid of the skateboard to the first side-×-angle position. The force transducer was then positioned at the side-×-angle position to measure the frontal-plane HA muscle torque. For the maximal HA muscle torque efforts (ie, the EMG data normalization trials), the subjects were instructed to “push as hard as possible into the pad until you hear a beep from the computer to stop.” The EMG and force data were sampled for three seconds after the voltage (ie, force) trace on the oscilloscope indicated a maximal and sustained level. During the sampling, the experimenter set the level of the maximal sustained voltage for later use as feedback. The subjects were told that they could use their nonetest limb in any way they desired for stabilization purposes. Subjects were instructed that they should request a retrial for any trial that they did not feel was a maximal effort.

Immediately following the maximal contractions, the subjects produced a contraction that was 20% of the maximal torque magnitude. The supine subjects were able to easily observe the dial setting on the feedback meter because it was placed directly in line with their field of vision. When the subject was able to maintain the 20% submaximal contraction, the force and EMG data from both hips were sampled for a four-second duration. Maximal and submaximal efforts were repeated for this initial side-×-angle condition following a 60-second rest. The sampling procedure was then repeated for the contralateral hip of this particular test angle, which completed the entire first part of the overall experiment. This entire procedure of part one was reiterated for each of the remaining five test angles.

Data Reduction and Analysis

Each HA muscle submaximal effort from which data were collected will be referred to as a NDI trial. After testing, the NDI for each trial was calculated by first normalizing the EMG activity (in millivolts) produced from each submaximal HA muscle torque effort to a percentage of the EMG activity produced during MVC. This EMG value was then divided by the absolute magnitude of associated submaximal torque (in newton-meters). Each subject performed a total of 24 NDI trials (ie, two sides × six test angles × two contractions). The two NDI trials for each side-×-angle condition were averaged to produce 12 NDI observations per subject. All subjects completed all trials, except for one subject who was not able to complete a 40-degree left HA muscle contraction because of muscle cramping.

The dependent measure in this experiment was the NDI observation of the HA muscles during a four-second submaximal isometric torque effort. A three-way analysis of variance (ANOVA) with a random factorial arrangement was used to statistically analyze the results. The three independent variables (or factors) of interest were subject, side, and hip angle, with 40, 2, and 6 levels within each factor, respectively. The factor subject was added to the statistical model to serve as a “blocking factor” to control for intersubject variability. The alpha level of acceptance in all statistical tests was set at .05.

Results

Table 1 contains the descriptive statistics for the HA muscles’ NDI for the variables of side, angle, and side-×-angle combination. Table 2 displays the ANOVA summary data for the variables subject, side, angle, and side-×-angle. A statistically significant main effect was demonstrated for the variables of subject and angle (p < .0001). Figure 3 shows the NDI sample means by angle, averaged for both sides. A Tukey’s post hoc test showed that all interangle NDI means were statistically different from one another (p < .05), except for the difference between the angles of −10 and 0 degrees of abduction and between 0 and 10 degrees of abduction.

The interaction between side and angle was not statistically significant. No statistically significant intra-angle differences between right and left hip NDIs existed, regardless of angle (p > .05). Figure 4 shows the NDI sample means as a function of hip abduction angle for both sides.

Discussion

We have chosen to define myoelectric efficiency as the reciprocal of the NDI. Relatively high NDIs would indicate low myoelectric efficiency because a relatively high neural activation was required for the production of a 20% submaximal torque. We have stated that a relatively high NDI represents a condition of active insufficiency. This efficiency concept of muscle has been used in other research.19,20

Neural Drive Index as a Function of Hip Abductor Muscle Length (Hip Angle)

Without regard to hip side, Figure 3 shows that NDI increased as the length of the HA muscles decreased. The NDI increased rapidly, particularly at hip angles greater than 10 degrees of abduction. We did expect the NDI to increase as the length of the muscle progressively shortened as a general response to the active insufficiency encountered. Increasing levels of EMG activity as a function of decreasing muscle length have been reported for maximal effort contractions for other muscles.21,22 In this study, the NDI was based on a ratio and therefore was sensitive to changes in the numerator (ie, EMG activity, expressed as % MVC) or the
denominator (ie, the absolute value of the 20% submaximal torque), or both. It is possible that the increases in NDI at shorter lengths did not reflect increases in neural drive. Rather, the EMG signal may have remained constant across muscle lengths, and the accelerating NDI may have reflected merely the decline in absolute torque, which we know to occur at shorter muscle lengths (Fig. 2). If these results were true, however, the NDI at the shorter muscle lengths should have increased in a more linear fashion, reflecting the linearity of the torque declines at shorter muscle lengths. On the contrary, the NDI-angle curve of Figure 3 showed a rapid nonlinear increase in NDI between 10 and 40 degrees, demonstrating that the EMG activity levels (ie, neural drive) increased concurrently with the absolute torque declines. We assume that excessive, and therefore relatively ineffective, actin and myosin overlap occurred at the shorter muscle lengths and that the CNS attempted to compensate for this active insufficiency by increasing the neural activation to the muscle.

Figure 5 is a comparative graph summarizing the maximal isometric right HA muscle torque response and the NDI as a function of muscle length for the same group of subjects.

In theory, a dramatic change in the HA muscles' internal moment arm (IMA) as a function of angle may cause the CNS to vary its output to the muscle. Hip angles that improve the mechanical advantage (ie, increase IMA) of the HA muscle mechanism would theoretically require a lower NDI. Olson et al demonstrated that the IMA of the HA muscles does change as a function of hip abduction angle (Fig. 6), but the change would probably not have a functional impact on the torque output. Note, however, that the IMA of the HA muscles appears to decrease at angles greater than 10 degrees, which, from a biomechanical perspective, may have slightly increased the neural drive.

At longer HA muscle lengths, which occur with angles between —10 and 10 degrees, relatively low NDIs were measured. These muscle lengths may create an actin and myosin overlap at the fiber level, which is conducive to high tension development, and therefore make neural compensation unnecessary. Perhaps the muscle spindle or the Golgi tendon organ provides afferent feedback to the CNS that assists in the regulation of required neural drive as a function of muscle length. Libet et al have shown that the EMG signal inhibition at long muscle lengths was partially abolished following tendon deafferentation.

### Table 1. Descriptive Data of Hip Abductor Muscle Neural Drive Index Means by Side, Angle, and Side-x-Angle Combinations for All Subjects (N = 40)

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<th>Maximum</th>
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*Right side had 240 observations, and left side had 239 observations (all observations are an average of two trials).

**80 observations per angle, except for 40-degree angle, which had 79.

***40 observations per side-x-angle combination, except for right side-40-degree combination, which had 39.

### Table 2. Analysis-of-Variance Summary of Neural Drive Index Results for All Subjects (N = 40)

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The parallel elastic fibers that surround and physically support the HA muscle fibers may be stretched and "loaded" near and in adduction, possibly creating a measurable "passive" (i.e., noncontractile) hip abduction torque. From a gross tissue perspective, the relative adducted hip angle places the iliotibial tract and fascia lata of the lateral thigh in a stretched position, which, according to research by Inman, adds significantly to the overall HA muscle torque. Recall that, unlike muscle torque production, passive torque generated by the elasticity in connective tissues does not directly produce EMG activity, which, in part, may explain the low EMG activity required at long muscle lengths.

Of interest, note that the high myoelectric efficiency (low NDI) of the HA muscles occurs at lengths that approximate muscle lengths used during the support phase of walking. Active and passive tensions from the HA muscles (and related fascia) are used during the single-limb support phase of gait to control the frontal plane stability of the pelvis. The stance hip (or hemipelvis) does adduct slightly during single-limb support, which would stretch and load the passive nonelectrogenic structures such as the iliotibial tract and fascia lata. Without this passive tension provided by the lateral hip fascia, virtually all the lateral tensions would be generated by relatively metabolically expensive active processes within the muscle fiber. This situation may significantly increase the energy demands on the body as a whole, especially considering the large number of steps taken throughout a given day. Because the hip normally adducts slightly throughout single-limb support during stance, the ipsilateral HA muscle fibers must contract eccentrically to help decelerate this adduction. Moritani et al report research that supports the popular notion that eccentric muscle contractions require less EMG activity than comparable concentric contractions. The HA muscle noncontractile and contractile tissues appear to be used during the stance phase of gait in a manner requiring minimal neural drive. This characteristic of the CNS is similar to the principle of "minimal muscle action" discussed by MacConaill and Basmajian. In a very general manner, the CNS appears to actively drive a postural muscle only when the peripheral noncontractile tissues are not capable of providing adequate passive resistance to perform the work demand. This parsimonious nature of the CNS would allow the connective tissues within and around the muscle to play an important role in sharing the antigravity work load and minimizing the demands for active muscle contraction. From a clinical perspective, chronic postural malalignment conditions may alter the passive length-tension relationships of certain supportive connective tissues, which may minimize their ability to produce passive tension and perform useful work. The responsibility for low-level tension production may shift toward the contractile muscle proteins and, in theory, would decrease the metabolic efficiency of a relatively simple act such as controlling postural sway during standing.

**Neural Drive Index as a Function of Hip Abductor Muscle Length (Hip Angle) and Hip Side**

We reasoned that the reported differences in the maximal torque-angle curves across hip sides would be associated with a disparity in the myoelectric efficiency (NDI) between hips at equivalent HA muscle lengths (hip angles). Based on the underlying reasoning of this experiment, this disparity would functionally express itself by a statistically significant interaction between the variables side and angle. That is, any statistically significant intra-angle NDI difference between right and left hips would be a function of the position of the hip angle. As reported previously, the interaction of side and angle was not statistically significant. Therefore, we accept the null hypothesis that the NDI of the HA muscles was statistically equivalent for both right and left hips. However,
we believe it reasonable to offer a possible explanation for the nature of the trends displayed in the side-specific NDI-hip abduction angle curves of Figure 4.

Because of the right pelvic adduction posture and the difference in the maximal torque-angle curves across sides, we hypothesized that the resting length of the right HA muscles would be biased toward adduction. We, therefore, expected that the shortened muscle length inherent to 40 degrees of abduction would "slacken" or predispose to active insufficiency to a greater extent in the right versus the left HA muscles. The plot of the sample means in Figure 4 demonstrates that the right HA muscles required greater NDI, or demonstrated less myoelectric efficiency, than the left HA muscles at 40 degrees of abduction. This high neural compensation measured specifically at 40 degrees of abduction on the right side suggests this to be the most slackened condition of the 12 experimental conditions. However, because of the lack of statistical significance of our data, we cannot state that the right side-minus-left side NDI observed at 40 degrees of abduction was greater than the value that may occur simply from random variation of the NDI unique to our subject group.

At the adducted (longer) muscle lengths, the right HA muscles would experience a length similar to that encountered in the pelvic adduction posture described by Kendall and McCreary. We hypothesized that the right HA muscles would demonstrate a lower NDI than the left HA muscles at 10 degrees of adduction because this position would approximate the "position of function" for this muscle group. If the resting length of the right HA muscles is biased toward adduction, we would logically predict that a submaximal torque produced in adduction would require less neural drive per level of submaximal torque than the left HA muscles. Minimal neural drive would be necessary to drive either the right or the left adducted muscles at a submaximal...
torque because passive tension would be provided by the lateral fascia. The right side, however, should demonstrate enhanced myoelectric efficiency, given that this muscle group’s protein structure has adapted to provide for optimal actin and myosin overlap at its position of function. As shown in Figure 4, the NDI was lower for the right side than for the left side at -10 degrees (as well as through 30°). The reader must be advised, however, that the right side-minus-left side differences in NDI at and near adduction were not statistically different; therefore, this discussion should be considered within this limitation.

We do feel there is practical (vs statistical) significance in the fact that the variables of side and angle interacted in the same predicted direction (ie, the right side-minus-left side differences in NDI at and near adduction were not statistically different; therefore, this discussion should be considered within this limitation. Neumann et al estimated that the length difference between the two HA muscle groups of their subjects was about 6%. This postural length asymmetry may have been an inadequate stimulus to induce a statistically significant, measurable neural compensation at the 20% torque level. Further research is needed using the NDI on other posturally elongated muscles, particularly with muscle length asymmetry more profound than that encountered in this study.

The best supporting evidence that the right HA muscles function at a longer (ie, more adducted) resting length than the left HA muscles in our subjects is the statistically significant side-specific disparity measured in the maximal torque-angle curves. The nature of the disparity in the NDI-angle curves between hip sides reported in this study also tends to support a posturally induced shift in right HA muscles’ resting length; however, the evidence was not statistically significant and therefore cannot be considered conclusive. Many clinical situations involve patients with muscle-length asymmetry secondary to posture or joint immobilization that is much more profound than that explored in this study. Clinicians should be aware of the likelihood that muscle does attempt to adapt to chronic length change, and the nature of such adaptations should always be kept in mind.

Conclusions

1. The NDI of the HA muscles increased rapidly as their length approached their shortened position. The high myoelectric efficiency exhibited by the HA muscles at the adducted hip angles reflects the functional role that the noncontractile lateral hip fascia serves in producing low-level passive hip abduction torques such as that used during walking.

2. The disparity in the NDI-abduction angle curves between right and left hips was not statistically significant. Evidence to support a posturally induced difference in the functional resting length of the two HA muscle groups rests on the statistically significant differences in the slope of the maximal torque-angle curves between sides. The nature of the trends of the NDI-hip abduction angle data between sides justifies additional research into the use and refinement of the NDI, particularly when attempting to evaluate specific posturally induced changes in muscle function.

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